PROSTHESIS USING MYOELECTRIC CONTROL

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In Partial Fulfilment of the Requirements
for the Degree of
MASTER OF TECHNOLOGY

by

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MY PARENTS

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R. Baliga

CERTIFICATE

This is to certify that the thesis entitled, "PROSTHESIS USING MYOELECTRIC CONTROL" has been carried out under our supervision and that it has not been submitted elsewhere for a degree.

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CONTENTS

ABSTRACT

A prosthesis is a device which is fitted to an amputee to replace the missing limb. A myoelectric control system is a man-machine system that utilizes command information encoded in the skeletal muscle. Various kinds of prosthetic devices are available designed to suit the needs of an amputee. In this project 'Prothesis using myoelectric control', an attempt has been made to design a myoelectric control system and to fabricate a particular prosthetic device, namely, an externally powered myoelectric hand for a below elbow amputee. The device designed is meant to enable an amputee to flex or extend the artificial fingers provided.

A general study of biochemistry and the nature of the myoelectric signals has been made. The design of the electrode to acquire the signals has been made on the basis of the detailed study of the signal characteristics. Electrodes with built-in difference amplifiers have been fabricated to acquire the extensor and the flexor signals, rejecting the common mode noise. A processor unit consisting of notch filters to reject line frequency pick-up, bandpass filters to reject unwanted frequencies and amplifiers to amplify the signals to the desired amplitudes, has been design and fabricated. The processed signals are rectified, average

and compared with certain preset threshold levels in the comparator section and the output of the comparator is given to the driver which excites the D.C. permanent magnet motor according to the command information.

In the comparator unit, an anticoincidence logic has been employed such that the output keeps the actuator idle if both the input d-c signals viz. flexor and extensor are either above the upper threshold or below the lower threshold. The actuator is made to flex if the d-c flexor signal is above the upper threshold and d-c extensor signal is below the lower threshold. The actuator extends the fingers if the d-c extensor signal is above the upper threshold and d-c flexor signal is below the lower threshold. To render the system immune to noise and jitter, comparators are designed to exhibit hysteresis.

A demonstration model of the actuator has been fabricated which converts the output of the motor into the movements of the artificial fingers. Laboratory testing of this device has been carried out and the feasibility of the scheme has been established, though, for practical application, the system has to be miniaturized.

CHAPTER 1

INTRODUCTION

Research and development into electrically operated limb prostheses for amputees, has progressed to such an extent that powered 'prosthetic' devices are now available for use on an experimental basis. The need for the development of such devices was motivated by the desire for an amputee to look normal and to function normally[1].

A prostheses[1] is the addition of a device added to the human body to replace the missing limb. The human being and the prosthetic device, then, constitute a man-machine system. When the device uses external power and the system is operated by means of feedback control, it comprises a cybernetic system. Such a system is indicated in Figure 1.1.

A myoelectric control system[2] (MEC system) is a man-machine system that utilizes command information encoded in the electric potential that accompanies contraction of the skeletal muscle. The essential elements of an MEC system are shown in Figure 1.2.

The voluntary contraction of a muscle fibre involves depolarization and repolarization of the cell mambranes and this involves ion movement. The resulting electric potentials

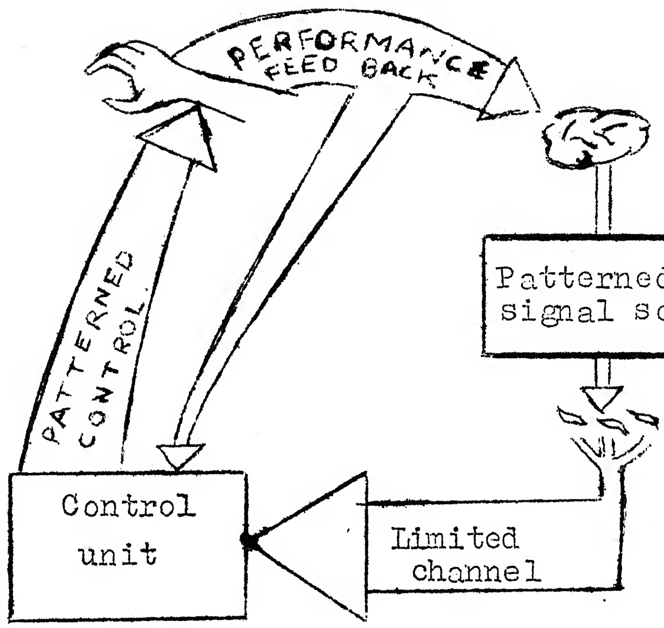


Figure 1.1: Man-machine system.

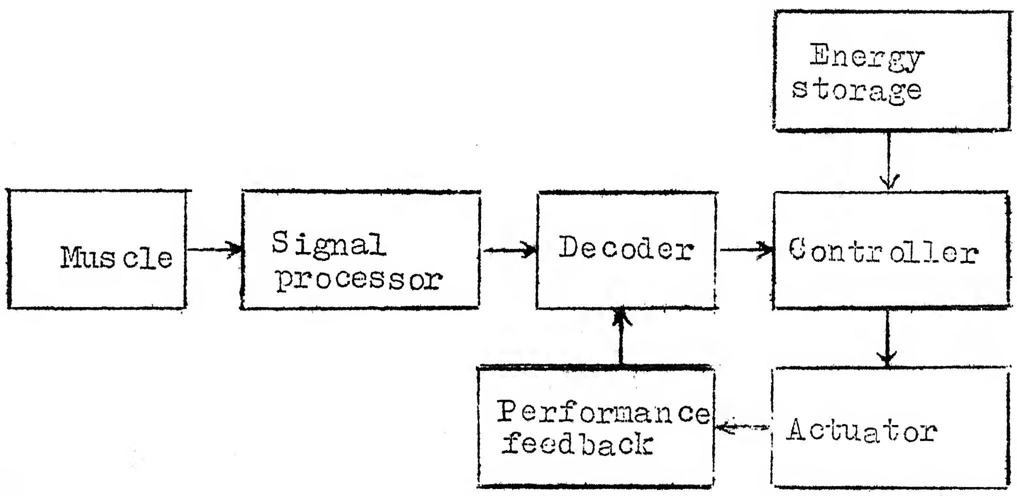


Figure 1.2: Block diagram of elementary myoelectric control system.

are called myoelectric signals[2]. These signals range upto about 100 micro-volts in amplitude.

The aim of this project was to make an attempt to construct a general MEC system which can be used, in particular, as a prosthetic device to control the movements of a specific artificial limb viz. the flexing and extending of an artificial hand. The prosthetic device was to be meant to enable a below-elbow amputee to function normally.

The general approach towards designing an MEC system consists of development of the following stages.

(i) Signal acquisition,

(ii) Signal processing,

(iii) Control system,

(iv) Output feedback (performance feedback).

The essential units of an MEC system are: Signal source, transducers, signal processor, output system and feedback receptors.

The signal source, essentially, is the myoelectric signal (MES) primed by the muscle movement. The basic function of the signal source is to translate a desire for action, burgeoned from the central nervous system, into a physiological signal. To achieve independent control of actions along the various degrees of freedom, it is necessary that the signal sources be independent[1].

Chapter 2 deals about the signal source, the characteristics of the myoelectric signals and some of the myoelectric controlled prosthetic systems developed by various research groups.

The role of a transducer is to transform the physiological energy associated with a signal source into a signal which reflects the "message" contained in the signal with a minimum of distortion and added noise. The myoelectrode is the most general and accepted form of transducer used in an MEC system[1]. A discussion on myoelectrodes, the material of the electrode and the design of the electrode is given in Chapter 3.

Signal processing is required when a signal such as an electro-myographic voltage has to be transformed to produce suitable input voltages for the control of the output system (actuators), which is analogous to the message contained in the original signal. The EMG signals have to be filtered, amplified, rectified, smoothed and stabilized so that this proportionality can be used. The complete processor unit and its design is discussed in Chapter 4.

The output system which includes the actual prosthetic device attached to the human body, basically, is the actuator. It usually consists of a motor, gears, levers etc. which translate the processed signal into the desired physical motion. The output system is described in Chapter 5.

In the last chapter, the project is summed up briefly and any further work that may be carried out in this field is described.

In an MEC system, especially in a prosthetic device, feedback constitutes an important unit. Man is endowed with excellent sensory feedback to his central nervous system which enables him to guide all parts of his limbs. In powered prosthetic device, a substitute for this natural feedback is necessitated. Vision forms the primary mode of feedback relied on in almost all functioning devices. Sound is also frequently relied on for feed-back information. Pressure, slippage, position, velocity and force have been used in various devices to enable automatic control of the actuator and to get a stable action with different degrees of success[1].

An amputee using a prosthetic device needs to be taught to use the device. This process is called Adaptive learning. Motivation and flexibility are of great importance here. It has been observed that children exhibited incredible learning ability with devices which are only marginal in their function[1].

An MEC prosthesis system enables an amputee to use the artificial limbs in near-normal fashion, though it is not an answer to all prosthetic problems. In certain cases, the control site might set the limit on the number

of independent signals that can be tapped, thus limiting the degrees of freedom presented by the device. In certain cases the message contained in the signal may not permit smooth functioning of the device, and the actuator may cause backlash. Even to this day, an MEC prosthetic device is not accepted with certainty.

In this project, the gross myoelectric signals (described in Chapters 2 and 3) were tapped from the muscles of the fore-arm, filtered and processed to drive the actuator which performs the flexing and extending of the artificial fingers. The performance feedback is basically visual.

CHAPTER 2

MES CHARACTERISTICS AND MES SYSTEMS

The tiny contractile fibres of a mammallian skeletal muscle originate the force responsible for any physical action. These fibres are origanized in groups called motor units, each motor unit having a single motor neuron to convey motor impulses from the central nervous system[2].

The contraction of a muscle fibre involves depolarization and repolarization of the cell membrane, and thus involves ion movement[2]. The electric potentials resulting can be measured easily and these potentials are called 'myoelectric signals'.

2.1 Motor Unit Action Potential

The most convenient starting point for a mathematical description of myoelectric signals is the motor unit action potential[2]. Figure 2.1.1 shows a typical single motor unit action potential observed with an intramuscular wire electrode using a processing system having a bandwidth of 8-1000 Hz[2].

To design any MEC system, one has to get well-acquainted with MEC signals and its characteristics. A collection of important extracts from the work done by various search groups[1-4] is presented below.

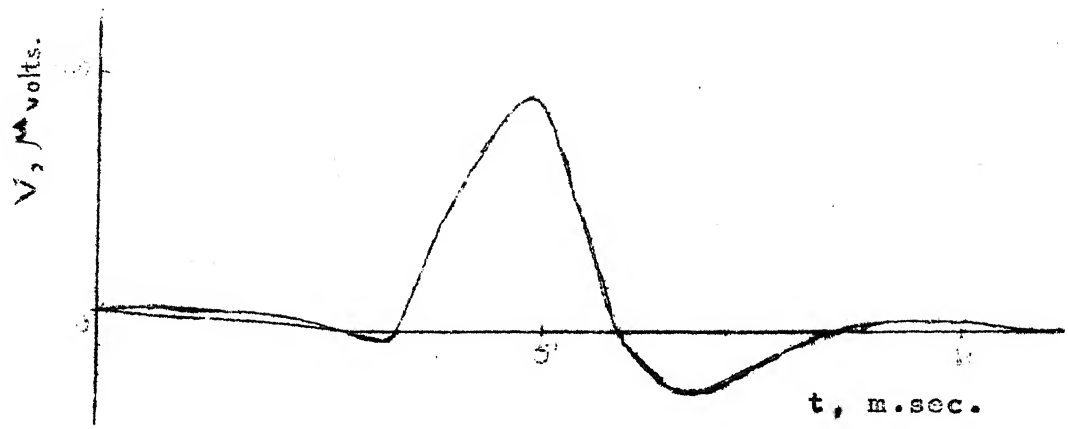


Figure 2.1.1: Typical single motor unit action potential.

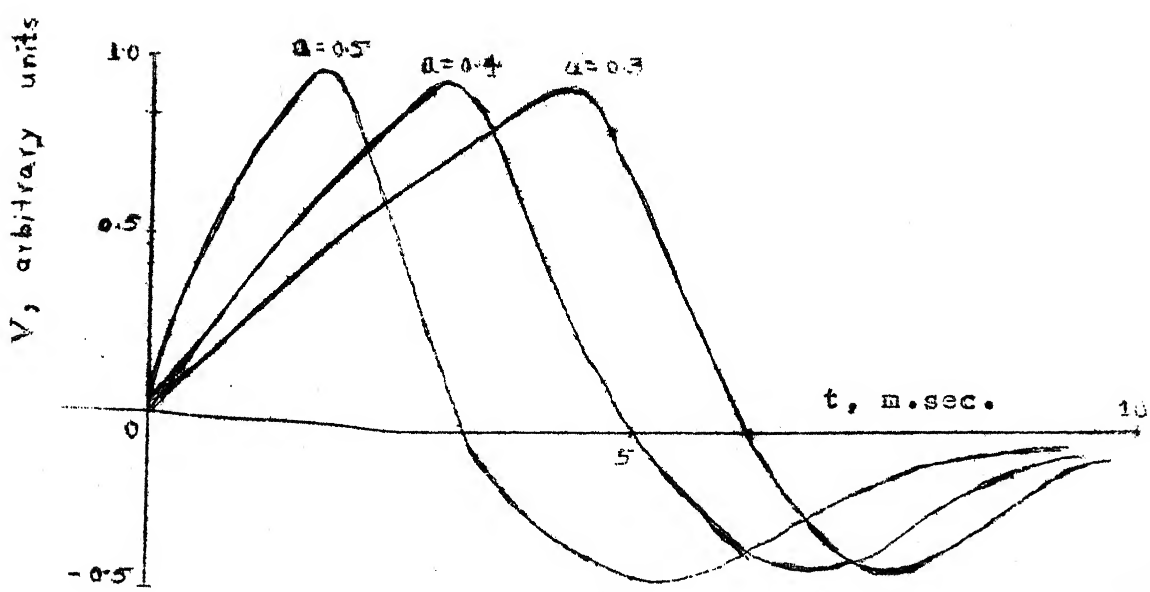


Figure 2.1.2: Model of single motor unit action potential.

$V = t(2-at)\exp(-at)$

In an MEC system, it is the gross myoelectric signal which is used. Hence, an exact description of each individual motor unit is not necessary. To explain the characteristics of gross myoelectric signal, which is a weighted sum of many trains of single motor unit potentials, one needs a model representing a single motor unit potential, having a shape that is amenable to mathematical manipulations. A convenient model which forms the basis for such an analysis is given by the equation:

$$V = Et(2-a\,t)\exp(-a\,t) \qquad (2.1.1)$$

where 'a' is a constant found to be between 0.4 and 0.45. This is plotted in Figure 2.1.2 for representative values of the co-efficient 'a' and for E = 1 unit.

The normal repetetive firing of a motor unit is not exactly periodic, but it can be modelled as having a rate that varies randomly about a mean firing rate.

A bipolar electrode may be considered as two monopolar electrodes spaced a distance 'd' along the direction of the muscle fibres. It has been observed that in a bipolar electrode, the signal at one electrode is identical to the signal at the other, except for a scale factor and a time shift of T = d/u where u is the velocity with which the wavefront propagates along the muscle fibres.

2.2 Gross Myoelectric Signals

Though the nature of transmission of electrical signal is similar for both nerves and the muscles, the physiological organization of muscle fibres is more complex. A muscle may comprise many thousands of fibres. The fibres are organized into subgroups (motor units) of upto a hundred fibres. In cross-section, the fibres of any one motor unit are interdigitated with the fibres of other motor units.

With the voluntary contraction of a muscle, the action potentials of the fibres of a particular motor unit are synchronous, but, the unit as a whole acts asynchronously with respect to the other motor units in the muscle.

The duration of the unit action potential has been determined to be around 10 m.sec. This varies considerably with different muscle and individuals.

The amplitude is dependent on fibre density and the number of muscle fibres of each motor unit. A pattern of gross myoelectric signal observed in the course of this project is shown in Figure 2.2.1.

The amplitude of the signal varies noticeably with the type of electrode used and the distance to the active muscle fibres. If surface skin electrodes are used, the signals due to various asynchronous neighbouring motor units also get added up. This superimposition of

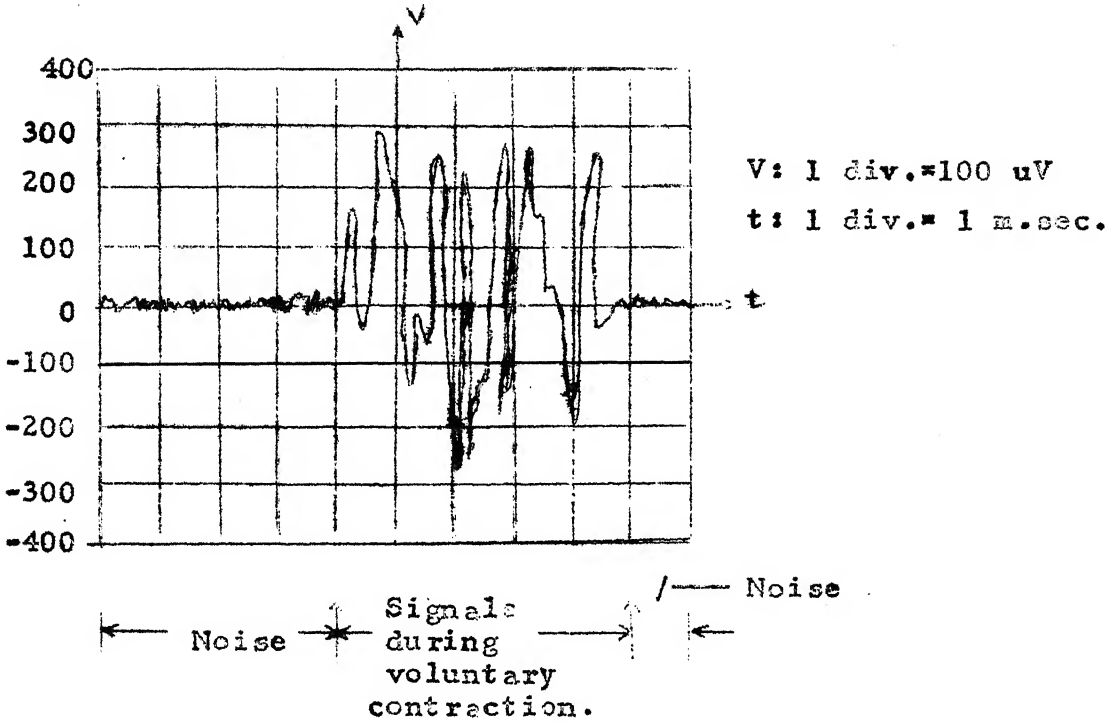


Figure 2.2.1: Gross-myoelectric signals observed.

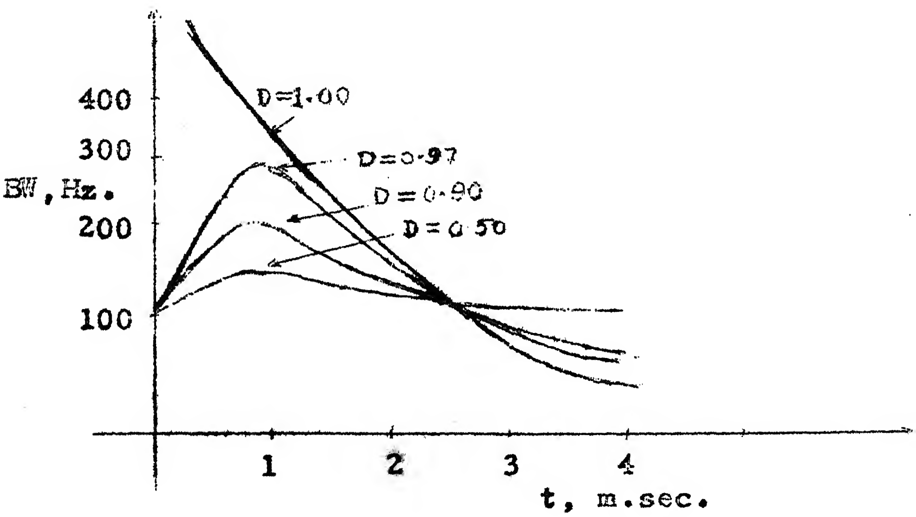


Figure 2.3.1: Bandwidth; Gross myoelectric signal (bipolar).

e signals from different asynchronous motor units has cacophonic effect and makes the detailed analysis of e signal difficult[3].

The muscle temperature has been found to have a egligible effect on the gross myoelectric signals and uscle fatigue, has been found to have no effect on the ignal amplitude at all[4].

.3 Signal Characteristics

a) Bandwidth: The bandwidth of the signals is seen to lepend on electrode spacing. Figure 2.3.1 shows a plot of bandwidth as a function of T where T = d/u, d being the spacing between the electrodes and u, the velocity of propagation of the wavefront.

(b) Power Spectrum: The power spectrum depends on the choice of the electrode. The activity picked up with the needle (intra-muscular) electrode contains components of higher frequencies than that obtained with skin electrodes. This is partly due to the change imposed on the motor unit action potential as it is transmitted through a tissue and picked up by a large electrode[4].

The power spectrum is also dependent on the position of the skin electrodes, the distance between the electrodes and the muscle fibres[4].

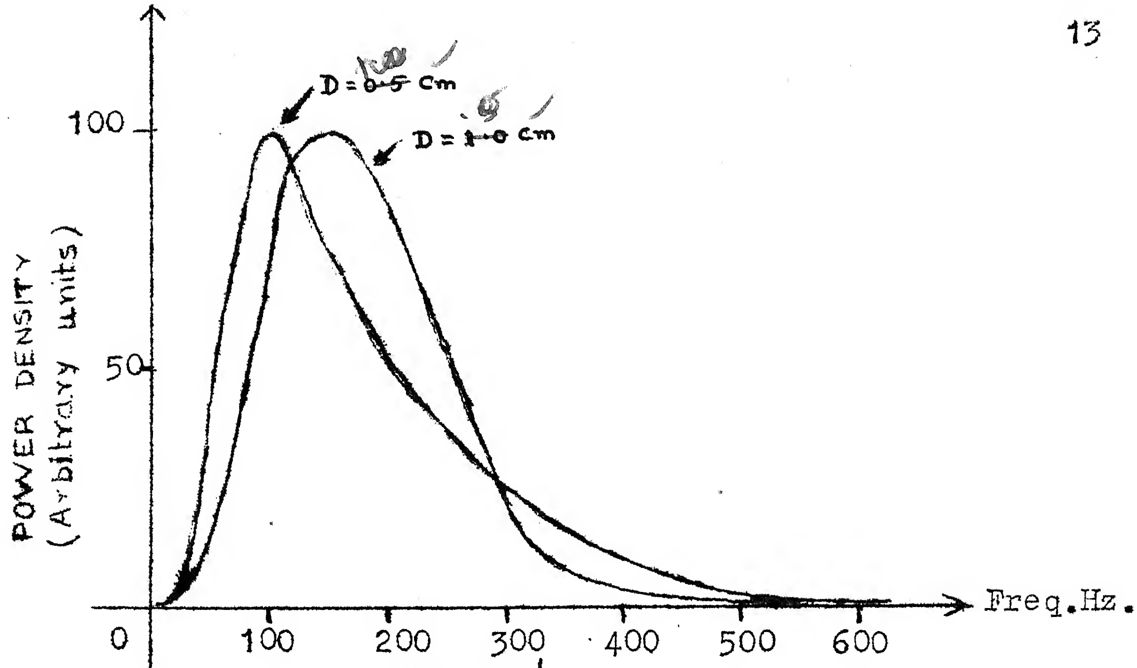


Figure 2.3.2: Measured Power density spectrum.

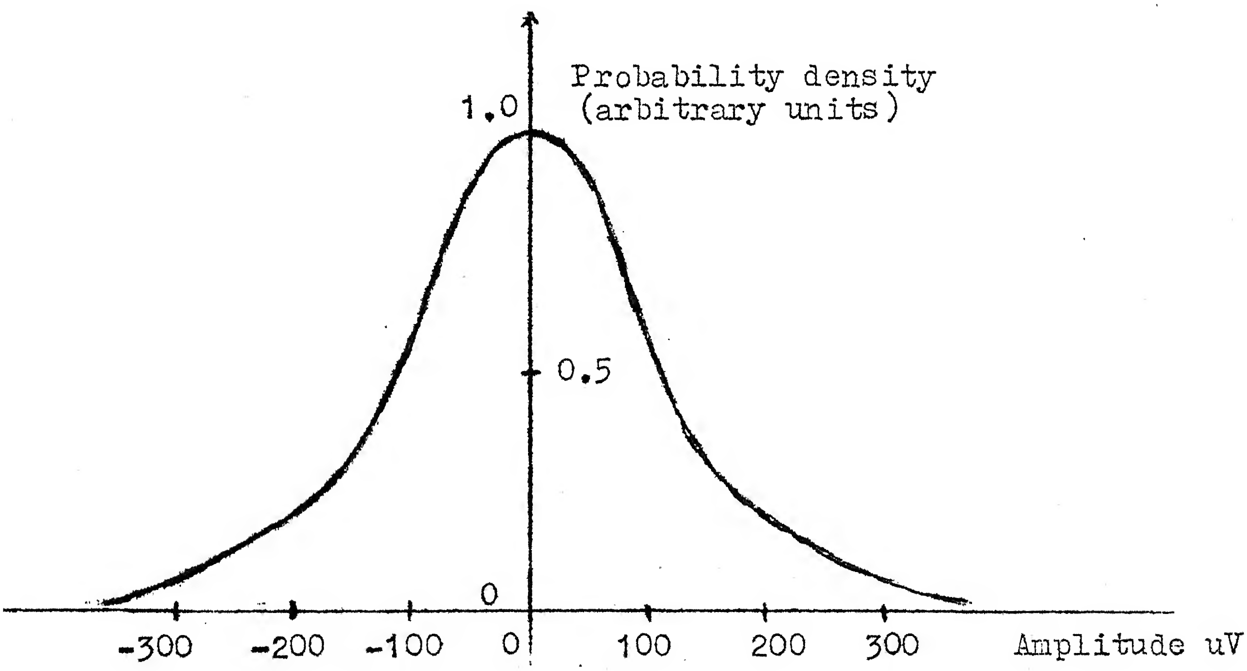


Figure 2.3.3: Measured amplitude probability density.

The power density spectrum of the bipolar signal is shown in Figure 2.3.2.

(c) Amplitude probability density function: As the electric fields in the various motor units differ in amplitude, phase, repetetive frequency and polarity, the individual motor unit potentials will add randomly at the electrode. At higher levels of contraction, more numbers of motor units start firing asynchronously. The myoelectric signals, therefore, assume the essential character of noise.

With the rise in the level of contraction, a rise in the potential duration and amplitude is effected. The amplitude probability density function of the gross myoelectric signal has been found to be approximately Gaussian[2], as expected for the summation of randomly occuring events. The approximationis more accurate for situations in which a large number of motor units are active. A typical amplitude probability density plot[2] is shown in Figure 2.3.3.

(d) Effect of electrodes on gross myoelectric signals

The intracellular potentials in skeletal muscle fibres are of the order of 100 mV, but the extracellular potentials are very much smaller, and peak amplitudes observed with intramuscular electrodes rarely exceed 10 mV.

When surface electrodes are employed, the preceeding comments on the myoelectric signal apply, but, with slight modifications. It is unusual, but not impossible, to observe clear single motor unit action potentials with the surface electrodes. The power density spectrum of the signal observed with surface electrodes is shifted slightly downward in frequency compared with that for intramuscular electrode. The effect of electrode spacing is not very noticeable at relatively large spacings, normally, greater than 2 cms.

One aspect of the myoelectric signal that is of particular interest in the design of practical control systems, is the available dynamic range. The signal during maximum voluntary contraction, is typically 300 μV.r.m.s. while a signal of about 100 μV r.m.s. corresponds to a comfortable, non-fatiguing contraction. A resting skeletal muscle is totally inactive apart from some noise, which has been attributed to various sources and does exceed the thermal noise calculated for the equivalent source impedance[2]. However, in most cases of clinical interest, myoelectric signals from nearby active muscles, rather than this background noise dictate the minimum usable signal for control purposes.

2.4 Signal Processing in Prosthetic Devices

From the details mentioned above, a powered prosthesis can be operated in two ways:

(i) On-off or threshold control,

and (ii) By proportional Control.

In On-off or threshold control, the signal is processed so as to operate a micro-switch. The amplitude of the signal is sensed and when the amplitude exceeds a predetermined threshold level, the switch is turned on which in turn operates the actuator. In the off-position of the switch, the actuator returns to the original position. This has an advantage that the electronic system is simple and cheap. The only disadvantage is that the function performed is predetermined and hence the amputee cannot modify the function according to his will.

In the proportional control, the signal is amplified, rectified, filtered and integrated so as to get a D.C. wave proportional to the signal amplitude and this is used to operate the actuator. Hence, the performance of the actuator would be proportional to the input signal. This has an advantage that the performance of the actuator can be easily controlled by theuser.

The control parameters chosen for the control of a prosthetic device should be easy to be utilized by the patient. As muscle tension can be controlled voluntarily, it is natural to seek a parameter directly correlated to the developed muscle tension. The most common description

of a myoelectric signal, so far, has been to state the 'Integrated mean value' or the 'Root mean square value' (r.m.s. value) of the amplitude[4]. The integrated mean value is easily obtained by amplifying, rectifying and filtering the myoelectric signal.

2.5 A Review of Existing Prosthetic Devices[1]

The Research Institute of Montreal (RIM) Myoelectric hand utilizes myoelectric signals from the extensor and flexor movements of the wrist. Stainless steel direct contact electrodes are used. The signal obtained is amplified, rectified and filtered to produce a D.C. voltage proportional to EMG power, to actuate the electric motor. The motor driven linkage opens and closes the forefinger in palmar prehension, as detailed by the input signal. Visual feedback is relied on and the sound of the motor provides the auditory cue. The movements achieved are natural, requiring the amputee, merely to 'think' of movements in a manner similar to that which he did before loosing the hand.

The Viennotone myelectric hand is the most successful commercial myoelectric hand. The principle of signal acquisition and processing is similar to the RIM hand. The additional advantage provided by this device is that the gear reduction system attached to the motor incorporates

a 'lock' which prevents torque from being transmitted back to the motor, permitting the device to maintain a prehensile force when applied. Visual and auditory feedback are relied upon.

The Otto Bock myoelectric hand manufactured in Germany is similar to the Vieunatone Hand. All the functional units except the output unit are very similar to the Vienunatone hand. The output unit has the capability of changing automatically, the gear ratio when the load increases. Thus, the hand closes initially at a relatively high rate but slows down with much increased prepension force as an object is grasped.

The National Institute for Aid to the Industrially Disabled (NIAID) myoelectric hand is also similar to the Viennotone hand, but for the fact that the speed reduction is accomplished somewhat differently, using a toothed belt which reduces the noise level.

The Automatic (Bel grade) Hand Prosthesis model, designed by Tomovic and associates, uses two signal sources to initiate the opening and closing of the hand. A third signal source is used to modulate force prehension. The signals are processed to actuate the motor. Mechanical switches are used to initiate remote control. Prehension is caused by tension on cables which automatically equalize

the force to the fingers. When a finger meets no resistance, it continues to curl up in flexion. Pressure-sensitive elements relying on pressure-sensitive paint are located at specific points in the fingers. On grasping an object, an appropriate sensor is touched and this initiates prehension until a predetermined maximum prehension force is applied. If the amputee desires greater prehension force, he may voluntarily override the system and increase the motor torque. The system acts as a relay feedback system when the release of the object is commanded, returning the whole mechanism to the initial position.

A Swedish research group has been working on the so-called SVEN Group Hand. The aim is to obtain at least four independent control signals from combination of EMG signals picked up by ten myoelectric contact elements.

The Warseda Hand developed in Japan, in addition to overall control by visual feedback, employs a pseudosensory pressure feedback to the patient. This is accomplished by means of a pressure transducer located in the thumb of the hand which is used to modulate a sensory signal to the forearm of the patient.

Apart from these models discussed above, a number of other ~~x~~research groups have been working in the field of prosthetic devices[5].

Despite the fact that some of the prosthetic devices mentioned above are commercially available, they are produced in limited quantities and are relatively expensive.

Good engineering has not revolutionized prothetics, but has rather improved it. With the present technology, emphasis on natural appearance tends to force compromise on function.

CHAPTER 3

SIGNAL ACQUISITION

Signal acquisition is of supreme importance in any biomedical instrumentation. The basic transducers used in myography are bioelectrodes[2,6,7]. Bioelectrodes are devices that transform biological and physiological phenomena into electrical currents, or conversely, generate such a phenomena from electrical currents (as in stimulation). Such devices make use of the fact that electrolytes in biological solutions and body tissue contain charged particles. The sole function of an electrode is to transfer charge between ionic solutions and metallic conductors.

The functionof the myoelectrode is to record electrical events resulting from the muscle fibres. These electrodes are quite different from the fluid bridge electrodes used for impedance measurement in membranes or chemical solutions. In this chapter, the discussions are restricted to myoelectrodes only. These electrodes are basically the same as used in electro-encephalography and electro-cardiography except that the myoelectrodes are smaller in size (in order to avoid interference from unwanted muscles.).

3.1 Charge and Charge Concentration[2,6]

To design any bioelectrode, a study of the origin and nature of the signals to be picked up by the corresponding electrode is essential.

In a human body, the movement of the muscle causes sodium or chlorine ions to move across the cell membrane[7]. In the body, a majority carrier is an ion such a sodium or intracellular potassium. Such ions in the body are quite mobile; and are present in large ~~conc~~ concentration so that when an electric current flows, they are in majority and carry most of the transferred charge. Minority carriers are ions in a very low concentration, not more than 1/20 the concentration of the largest majority carrier. Minority ions do not normally carry much of the total transferred charge.

Ions interface between metals and electrolytes and across membranes. They generate potentials independently when they are made to move. This potential is determined by the equilibrium between the opposing forces of diffusion due to existing concentration gradient, and electrostatic repulsion, due to the build-up of depletion layers.

The potential developed across a membrane due to a particular ion through the membrane is given by the Nernst equation[1,2,8] and can be stated as

$$E = \frac{-RT}{nF} \ln \frac{C_1 f_1}{C_2 f_2} \qquad (3.1.1)$$

where R = the gas constant (8.315×10^7 ergs per mole per degree Kelvin)

T = absolute temperature in degrees Kelvin

n = the valence of the ion

F = Faraday constant (96,500 Coulombs)

C_1, C_2 = the two concentrations of the ion

f_1, f_2 = the respective activity co-efficient of the ions on the two sides of the membrane.

This is based on an assumption that the solution is weak so that there is enough free space between ions for free movement. This is true for human body which contains 0.9% by weight of sodium chloride in the normal saline.

3.2 Liquid Junction Potential

When a membrane separates two different liquids, each of the different ions present on either side, will contribute its Nernst potential to develop a net potential difference across the membrane. One has to note that the mobilities of the different ions as they permeate through the membrane will, in general be different, e.g. in sodium chloride solution; sodium ion has less mobility due to its hydration and forms sodium hydroxide while chlorine ion permeates much more freely. This potential ranges upto 50 mV, depending upon the concentrations.

The junction potential also contributes to the bioelectric potential in the skin. The epidermis of the skin consists of a dead corneal layer. Beneath this, lies the dermis which consists of cells and blood vessels.

3.3 Motion Artefact

When surface electrodes are used, the relative motion of the skin with respect to the electrode causes sudden changes in potential because interfaces are being disturbed. Most of the voltage drop at a metal-electrolyte interface occurs in the inner layers of ions, The voltage then drops exponentially with distance until it reaches zero at the outer boundary. This potential depends upon the fact that the ions would have redistributed themselves.

In case of mechanical agitation of the interface, because of these redistributed ions, the potential distribution assumes a different shape, leading to the large potential change. The large voltage change at the electrode surface due to a mechanical disturbance, is an electrokinetic effect. The space-charge cloud gets washed away and it takes time for it to come back. Motion artefact can be minimized by mechanically stabilizing the interface area. Electrode

jelly is generally used to accomplish this. This jelly contains quartz or glass granules, which, when rubbed into the skin, breaks the epidermis and enables a contact between the dermis and the electrode[2]. The jelly also prevents movement artefact by acting as an adhesive agent.

3.4 Electrode Material

Nearly all myeelectric control systems employ surface electrodes. The system requirements are stability, reliability, low impedance electrical connection free from fluctuations of contact potential due to motion and small electrode area.

The choice of electrode material is limited by the physical toxicity and mechanical strength. The most commonly used material are the noble metals, platinum, silver, gold. Stainless steel, German silver and tantalum are also used sometimes.

Platinum is generally preferred as it possesses good electrical conductivity, is inert to the body and if alloyed with a small percentage of iridium, is rendered strong. Platinum is corrosion-resistant. This is generally used as intramuscular electrode.

Silver is a good conductor of electricity but it is mechanically soft and oxidizes easily, it is also quite prone to corrosion. Generally, silver electrodes with gold plating are used to overcome this drawback.

Surgical grade stainless steel is generally considered a satisfactory electrode material and may be used either as an intramuscular electrode or surface electrode. It is mechanically strong and resistant to corrosion and is non-toxic. It has an additional advantage of being economical relative to the noble metals[6].

When silver is employed, the contact impedance between skin and electrode can be reduced further by providing an electrolytic (silver chloride) interface between skin and the electrode. The silver chloride acts as a precipitate as it is sparingly soluble. It provides an environment of saturated silver chloride solution at the surface of the electrode. The silver chloride layer, thus, basically achieves a chemical path to communicate (interchange charge) with the sodium chloride environment of the body via chloride ions and interchange charge with the silver electrode via silver ions[2].

The use of Ag-Agcl electrode requires a thorough cleaning of silver layer and applying a paste of silver chloride. This nuisance shouldered by the patient is offset by the fact that one may use a lower input impedance system which inturn reduces the susceptibility to external electrical interference.

Some research groups[2] have been able to produce an intimate socket fit so as to permit the use of simple stainless steel electrodes without excessive movement artefact.

3.5 Design Considerations for the Electrodes

A surface electrode placed on the skin can be assumed to have the equivalent circuit[7,8] shown in Figure 3.5.1. However, the values of the resistance and capacitance tend to change with humidity. Hence to buffer this change of impedance, and to cause least attenuation of the amplitude of the signal picked up by the electrode, a buffer unit has to be included. This also improves the signal to noise ratio.

However, the lead from the amplifier to the electrode has to be screened to avoid electrical interference from external source. This lead would have a finite capacitance with respect to earths which would attenuate the signals. Now, if a two-electrode system is employed to avoid common mode noise (which is done using a difference amplifier with a high CMRR), then this lead capacitance could degrade CMRR of the difference amplifier. To avoid this, it has been suggested[9] to make the entire signal acquisition unit, as an integral part of the electrode. Using miniature I.C. components, it is possible to construct amplifiers with low input capacitance so as to record the EMG signals with least amount of attenuation and distortion.

The signal acquisition unit is shown in Figure 3.5.2. The requirements of the amplifiers discussed above necessitates

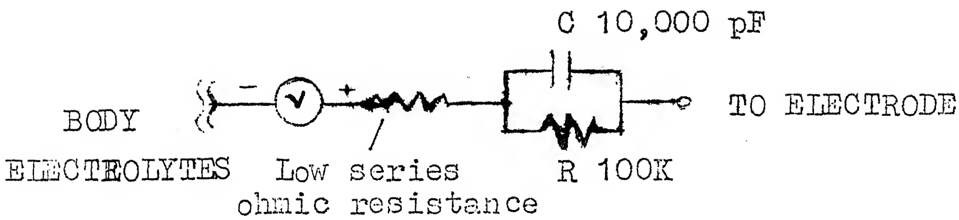


Figure 3.5.1:Equivalent Circuit presented by the body to the electrode.

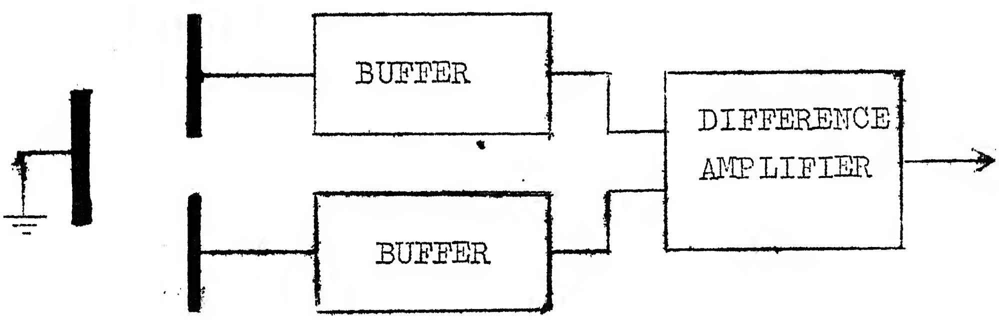


Figure 3.5.2: Signal Acquisition Unit.

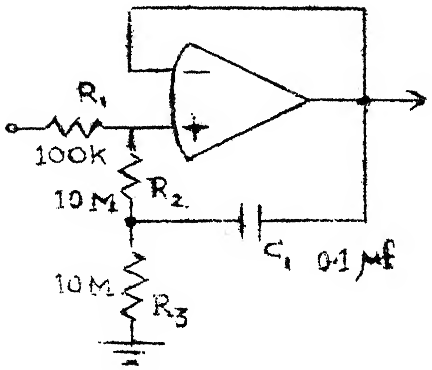
the use of high input impedance operational amplifiers (opamp) which require very low input bias currents, low input error currents, low drift and offset voltage and a fairly consistent operation over a large temperature range of -50°C to 100°C.

MOSFET input Opamps, having high input impedance and a few picoamperes bias currents are ideal but they are very costly and are quite prone to damage and requires careful handling. Another disadvantage with these Opamps is that the leakage currents though low at low temperature doubles for every 10°C rise in temperature.

The 'LM 308 A'[10] Opamps utilize matched super-beta bipolar input transistors and are thus capable of extremely high current gains. The input bias currents are of the order of 7 nA and input impedance, 40M-ohms. The super gain input transistors match much better than FETs and possess an input offset voltage of 0.5 mV (max.). This has a minimum CMRR of 96 dB, typically 100 dB. The worst case drift rate is 3 μV/°c.

3.6 Design of Signal Acquisition Unit

The design requirements of the Buffers demand a high input impedance, low output impedance negligible current to be drawn from the electrodes. This essentially dictates bootstrapped input unity gain amplifier. This is shown in Figure 3.6.1.

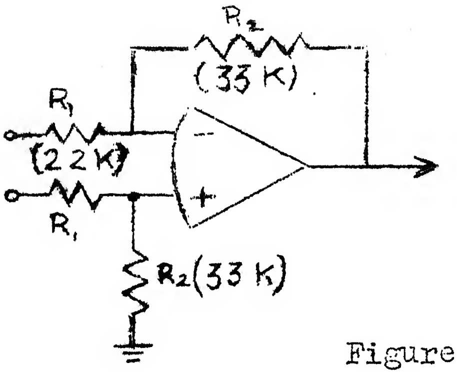


$$Z_{in} = R_2 + R_3 + (R_2 R_3 / X_1)$$

$$X_1 = 1/jwC_1$$

OPAMP: LM308A

Figure 3.6.1 Buffer.



$$V_o = \frac{(R_2)(V_2 - V_1)}{R_1}$$

OPAMP: LM308A

Figure 3.6.2: Difference Amplifier.

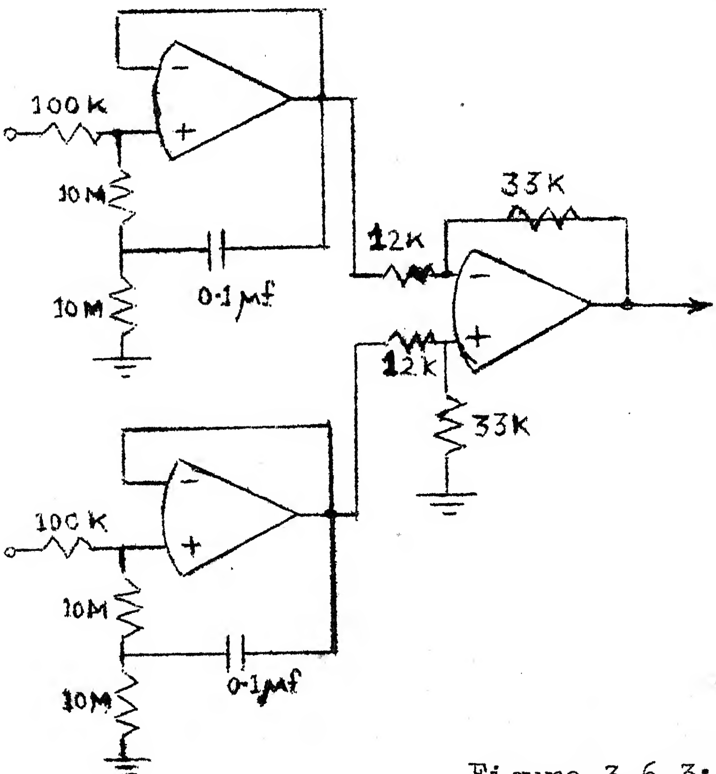


Figure 3.6.3: The circuit of the complete signal acquition unit.

The capacitance C_1 should be chosen large enough to act as a short circuit, at the lowest frequency of interest. The power spectrum of EMG signals shown in Figure 2.3.2 of Chapter 1 indicates that the bandwidth of interest lies from around 20 Hz to 2 KHz. Further at 50, Hz, a significant amount of interference is caused because of the proximity of power lines and hence it is preferable to reject all the frequencies upto and including 50 Hz. Hence 60 Hz may be considered as the lowest frequency of interest (giving a sufficient margin of 10 Hz for the drop in the frequency characteristics of the filter.).

The resistor R_1 is used to damp the variation in the source resistance R_s which can be considered to be of the order of 100K.[7] Assuming a tolerance of about 20% in R_s (worst case), if R_1 to of about 100K with 5% tolerance on its value, is chosen, then, the effective series resistance R_s+R_1 will have around 10% fluctuation. Larger values of R_1 will demand a much higher input impedance of the buffer to prevent attenuation.

If the attenuation of the signal is to be negligible, then, the input impedance Z_{in} will have to be of the order of 1000 M-ohm. Z_{in} of the buffer stage shown in Figure 3.6.1 is given by

$$Z_{in} = R_2+R_3+(R_2R_3/X_1) \qquad (3.6.1)$$

(assuming the large signal voltage gain of the Opamp to be very large; for LM308A, it is of the order of 300,000), where $X_1 = 1/jwC_1$.

The limit on R_2 and R_3 is set by the D.C. off-set voltage due to unequal bias currents that may be permitted. (If we choose to incorporate a gain of about 2.5 in the difference amplifier (as higher gains before filtering out noise would degrade signal to noise ratio.) and we decouple the D.C. component after the signal acquisition stage, then, an offset of about 5V may be permitted, as the only limitation is the possibility of saturating the Opamp at peak signal swings.

Since the biascurrent requirement is 10 nA(worst (case),

$$10^{-8} x(R_2+R_3)x2.5 \ll 5V$$

Therefore,

$$R_2 + R_3 < 200 \text{ M-ohm}$$

R_2 and R_3 are selected to be 10 M-ohm each as that gives a sufficiently high Z_{in}.

At a frequency of 60 Hz. (lowest frequency of interest),

$$|X_1| \ll R_2$$

Therefore $wC_1 \gg 10^{-7}$

C_1 = 0.1 uF gives $|X|_{f=60\ Hz} \cong$ 30K-ohms which is far small when compared to R_2.

This combination of R_2, R_3 and C_1 gives a Z_{in} of the order of 1000 M-ohms.

The difference amplifier circuit is shown in Figure 3.6.2. The output of this stage is given by

$$V_o = (\quad R_2)(V_2 - V_1)/R_1 \qquad (3.6.2)$$

Since, a limit of 5V on the amplified offset voltage of the buffer stages has been set, steps are to be taken to keep the DC offset voltage due to bias current of difference amplifier as low as possible. A limit of 0.1V may be assumed. Then, for the worst case input bias current of 10 nA, the resulting offset voltage at the output is given by $(R_2 x 10 x 10^{-9}/1.4)$ volts. Hence, a value of R_2 equal to 33K is arrived at.

The gain requirement of 2.5 yields a value of R_1 equal to 12K.

The complete signal acquisition stage is shown in Figure 3.6.3.

Stainless steel studs have been used as the surface electrode. The electrodes have been placed at a distance of approximately 2 cms. The entire signal acquisition unit

has been fabricated on a small elliptical card and steel studs fixed to it.

The assembled electrode is shown in Figure 3.6.4. The electrode can be strapped on to the hand using a rubber strap. Two such electrodes have been made; one, to acquire the flexor signals and the other, to acquire the extensor signals.

FIGURE 3·6·4 : THE ELECTRODES.

CHAPTER 4

SIGNAL PROCESSING

A method to acquire the signal has been described in the preceding chapter. After the signal is acquired, it needs to be processed according to the mode (ON-OFF or proportional control) of operation required of the system. To improve the signal to noise ratio, the signal has to be filtered and then amplified. Before the signal is further processed, the mode of control has to be decided upon.

The mode of control that has been adopted in this project is as follows. If the amplified signal is averaged after half wave rectification, a d-c voltage proportional to the signal is obtained. Two d-c outputs are made available, one proportional to the extensor input and the other, proportional to the flexor input. If threshold values for these outputs are fixed such that if the extensor (flexor) d-c voltage exceeds a fixed threshold value and the flexor (extensor) d-c value is below another fixed threshold value, then, the actuator operates in order to extend (flex) the 'fingers'. This scheme has an advantage that, under no signal condition, the system is insensitive to noise as long as the d-c output generated by noise has a magnitude less than the upper threshold.

further, the system is thereby made capable of clearly distinguishing the flexor and the extensor outputs. In doing this, the proportionality between the input signal and the output of the system is lost, but at the same time, reliability of the system is improved. The visual feedback that exists in the operation of the system ensures stability of the system.

The comparator outputs are then used to excite a d-c permanent magnet motor through a driver circuit.

4.1 Design Considerations for the Processing Unit

The output signals from the signal acquisition units are generally of the order of 1 mV and varies with individuals, the condition of the muscle stump and the type of electrode used. The required voltage gain is therefore of the order of 10^4. In order to prevent oscillations which might ensue due to this large gain, the gain should be distributed over the entire unit. The block diagram of the processor unit is shown in Figure 4.1.1.

The notch filters tuned at 50 Hz are introduced to filter out line frequency pickup. High-pass and low-pass filters are designed to permit only the frequencies of interest viz. 10 Hz to 1.5 KHz. The

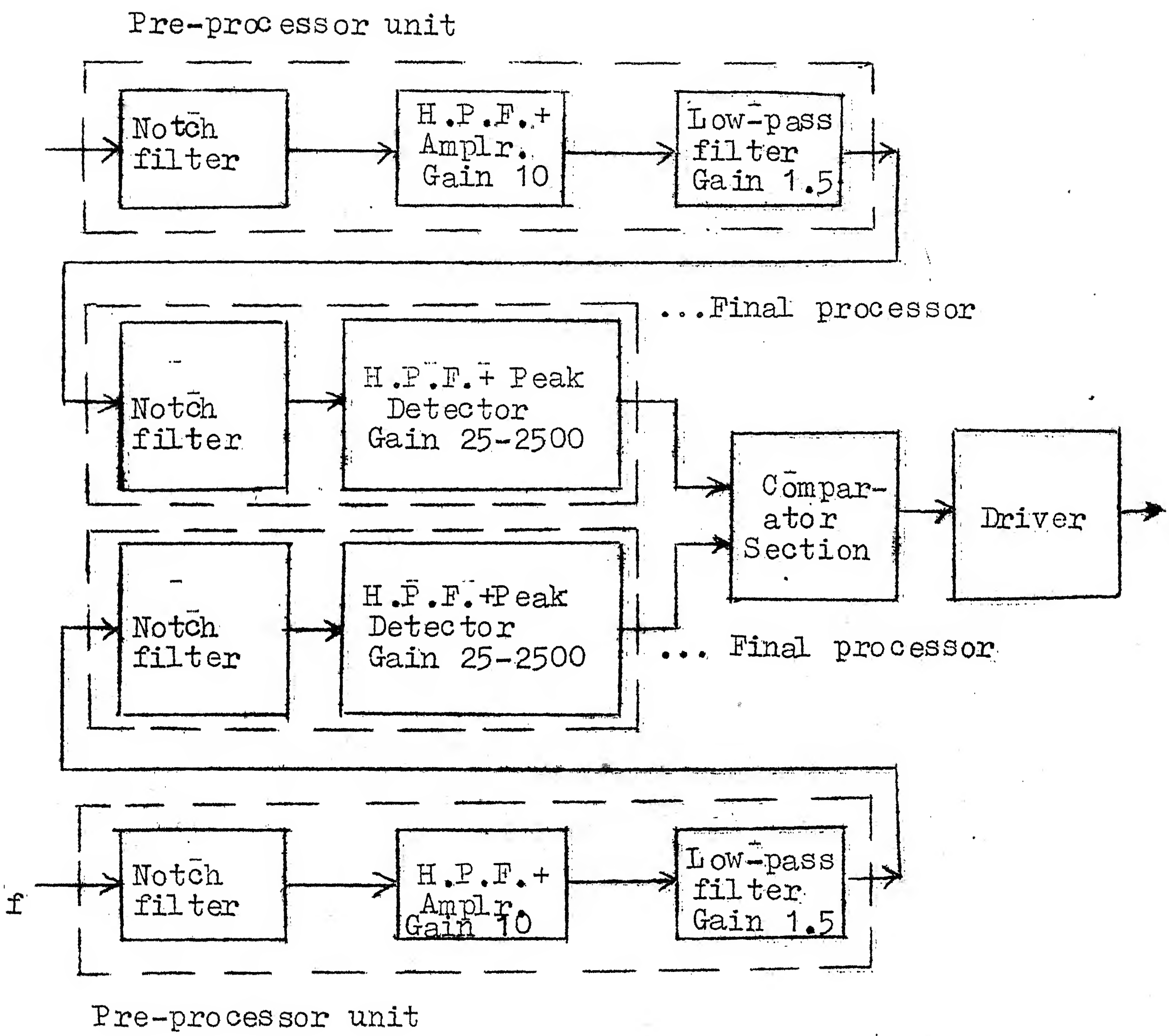


Figure 4.1.1: Block diagram of the processor unit.

The half wave rectifier is provided with a gain of 25 and the peak detector circuit is provided with a variable gain that can be manually varied from 1 to about 100. The filter stages are also provided with appropriate gains. This gives us a d-c output which is proportional to the average value of the input signal. This d-c output is then compared with the threshold values set by the hysteresis comparator. The output of the comparator goes to the driver. The driver circuit excites the motor as dictated by the outputs of the comparators of the two channels.

Care is taken to de-couple the d-c offset whenever necessary by introducing coupling capacitors indicated as HPF. Since low bias currents or high input impedance is not required of the opamps, μA741 opamps are used.

4.2 Design of the notch filter

The circuit diagram of the notch filter is shown in Figure 4.2.1(a) and its equivalent circuit is shown in Figure 4.2.1(b). The principle of operation is similar to that of a difference amplifier. At the notch frequency (50 Hz in this case), the R-L-C series arm gets tuned and the input appears as common mode signal and thus get attenuated. The normalized frequency response is given by

$$H(s) = \frac{(s^2+1)}{s^2+1+s\left(\frac{A_{Vo}}{Q}\right)} \qquad (4.2.1)$$

where

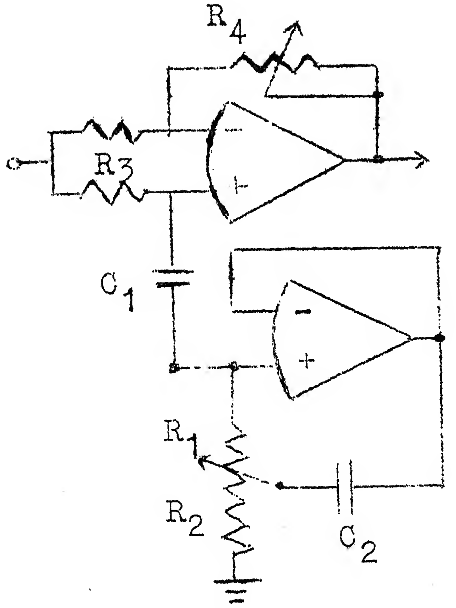
$$s = jf/f_N \ ,$$

$$Q = \left(\frac{2\pi f_N \ C_2 \ R_1 \ R_2}{R_1 + R_2}\right)$$

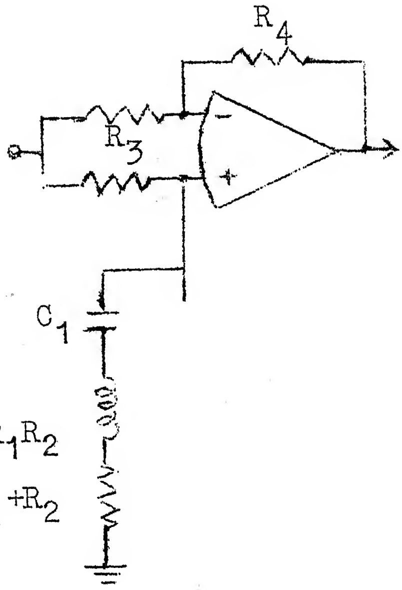
and
$$f_N = \frac{1}{2\pi\sqrt{C_1 \ C_2 \ R_1 \ R_2}}$$

Since the line frequency may vary slightly around 50 Hz, a low Q is desired; but if Q is too low, frequencies in the vicinity of 50 Hz get attenuated. Hence a Q of 1 to 3 is taken as a fair compromise between the two limits. Fixing C_1 and C_2 as some convenient values, R_1 and R_2 may be a trim pot, with centre tap such that the product R_1R_2 tunes the filter at the desired frequency and the maximum resistance between the two fixed terminals decides the Q. It is desired to have R_4 as another variable resistance of maximum resistance slightly more than the sum (R_1+R_2) so as to enhance fine adjustment to match the resistance of the R-L-C arm at the notch frequency.

The values arrived at, to tune the filter to reject 50 Hz input with a Q of about 2.0, are indicated in the figure. To minimize the shift of the notch frequency with temperature polycarbonate capacitors are used. The actual rejection achieved by this circuit to an input test signal of 50 Hz has been measured and found to be 35 dB.



(a) Notch filter



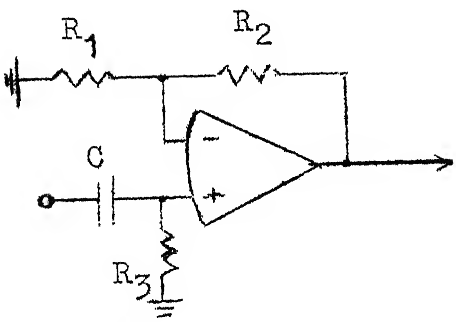
$L_1 = C_2 R_1 R_2$

$R = R_1 + R_2$

(b) Equivalent diagram

f_0 = 50 Hz
$R_1 + R_2$ = 100K(pot)
R_3 = 18K
R_4 = 200K(var.)
C_1 = 0.02 uF
C_2 = 0.47 uF
Q = 2.5

Figure 4.2.1: Notch filter.



R_1 = 100K
R_2 = 1 M
R_3 = 100K
A_V = 10
C = 0.47 uF

Figure 4.3.1: Non-inverting amplifier.

4.3 Design of the Amplifier Section and the HPF

The circuit of the simple non-inverting amplifier is shown in Figure 4.3.1. Before this amplifier, care has to be taken to avoid d-c coupling of the input to eliminate offsets accumulated from previous stages. Hence a large capacitor is provided. The combination of C and R_3 then acts as a high-pass filter. The cut-off has to be about 10 Hz, just enough to eliminate low frequency drift.

Therefore,

$$\frac{1}{2\pi R_3\ C} \simeq 10\ \text{Hz}$$

The parallel combination of R_1 and R_2 should be equal to R_3 in order to minimize the d-c offset due to unequal bias current. Further, the gain requirement dictates $(R_1+R_2)/R_1 = 10$.

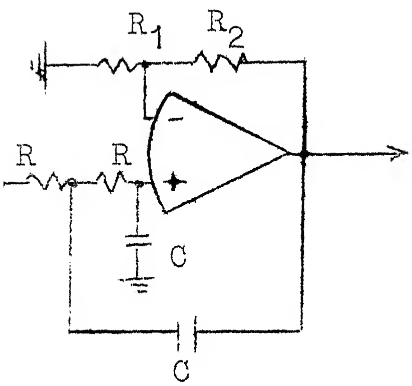
The values of R_1, R_2 and R_3 arrived at, are shown in the figure.

4.4 Design of the Low-Pass Section

A low pass filter has to be used to eliminate frequencies above the desired signal band viz. 1.5 KHz. For a second order filter,

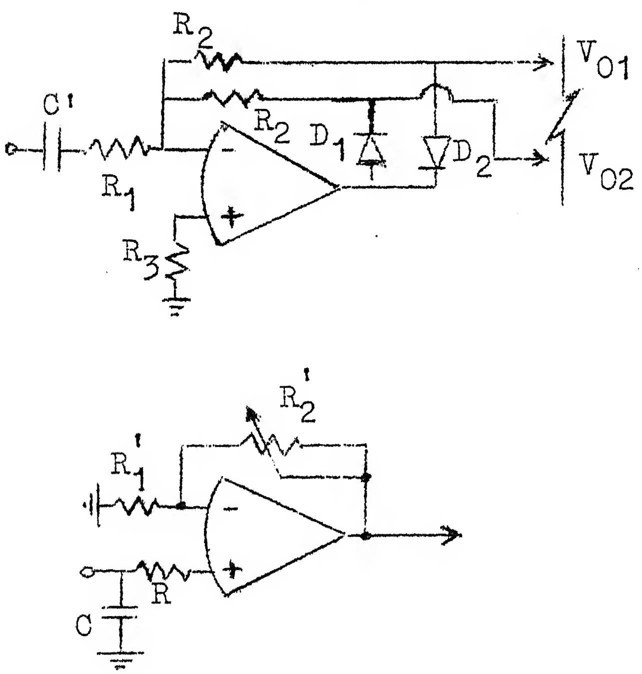
$$A_V(s) = A_{Vo}/((s/w_o)^2 + \alpha(s/w_o) + 1) \qquad (4.4.1)$$

For maximallly flat frequency response, $\alpha = \sqrt{2}$.



$$f_C = 1/2 \quad RC$$
$$A_V = 1.545$$
$$R_1 = 33K$$
$$R_2 = 18K$$
$$R = 2.2K$$
$$C = 0.01\ uF$$

Figure 4.4.1: Maximally flat low-pass filter.



Diodes D_1 & D_2 are low leakage SH100 diodes.

$$R_1 = 33K$$
$$R_2 = 820K$$
$$R_3 = 33K$$
$$R_1' = 10K$$
$$R_2' = 1\ M$$
$$R = 10K$$
$$C = 0.47\ uF$$
$$C' = 0.47\ uF$$

Figure 4.5.1: The peak detector section.

The circuit of the low-pass section is shown in Figure 4.4.1. The resulting expressions for A_{Vo} and w_o are as follows:

$$\alpha = 3 - A_{Vo}, \quad A_{Vo} = (R_1 + R_2)/R_1$$

and $w_o = 1/RC$

For a cut-off at about 1.5 KHz, the values of R_1, R_2, R and C arrived at, are shown in the figure. This section contributes to a gain of about 1.6.

4.5 Design of HPF and the Peak Detector Circuit

The circuit of the HPF and the peak detector circuit is shown in Figure 4.5.1. If V_i goes positive, D_1 is ON, D_2 is OFF and the circuit behaves as an inverting opamp. For negative going input, D_1 is ON and D_2, OFF. Hence the output V_o gives inverted, amplified version of the positive half of V_i and V_{02} gives inverted amplified version of the negative half of V_i.

A gain of about 25 is desired of this section. Hence $(R_2+R_1)/R_1 = 25$. R_3 is selected so as to minimize d-c-off-set due to bias currents. A-c coupling is needed before this section to prevent amplification of offsets carried on from the earlier stages. The values of R_1,R_2 and R_3 arrived at, are shown in the figure.

A simple low-pass R-C section is cascaded with this unit to get a peak detector. The charging time constant of this is the product of the forward resistance of the diode together with the resistance R_o, and the capacitance C. The discharging time constant is R_2C.

The values of R_2 and C are selected suitably. A further gain of 1 to 100 is needed to amplify the average value of the signals to the required value. R_1' and R_2' are selected to provide this gain R_2' is a variable potentiometer mounted on the side panel to provide for gain according to personnal need.

4.6 Design of the Comparator Unit

The purpose of the comparator has been explained at the beginning of the chapter. According to the mode of control decided upon earlier, the region of operation gets defined as shown in Figure 4.6.1. V_{IE} and V_{IF} are the average values of the amplified extensor and flexor signals. In the region A, corresponding to the relaxed condition of the hand, both V_{IE} and V_{IE} would be below V_{IL}. When the hand is flexed, V_{IE} would be below V_{IL} and V_{IF}, above V_{IH}, the operation being in region B. In region C, when the hand is extended, the input V_{IE} would be above V_{IH} and V_{IF} below V_{IL}. In the region D, both the signals V_{IE} and V_{IF} are above their respective upper thresholds.

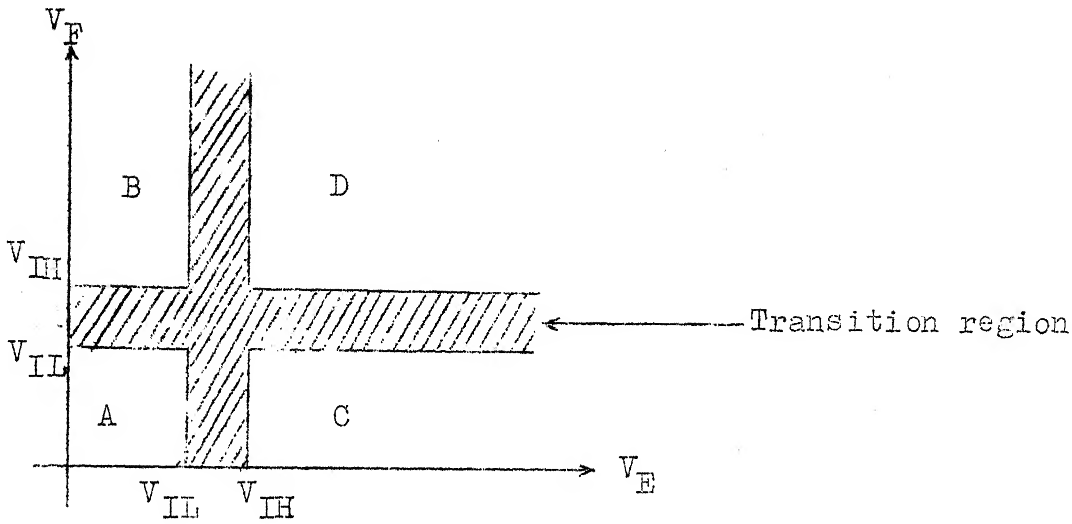


Figure 4.6.1: Actual region of operation.

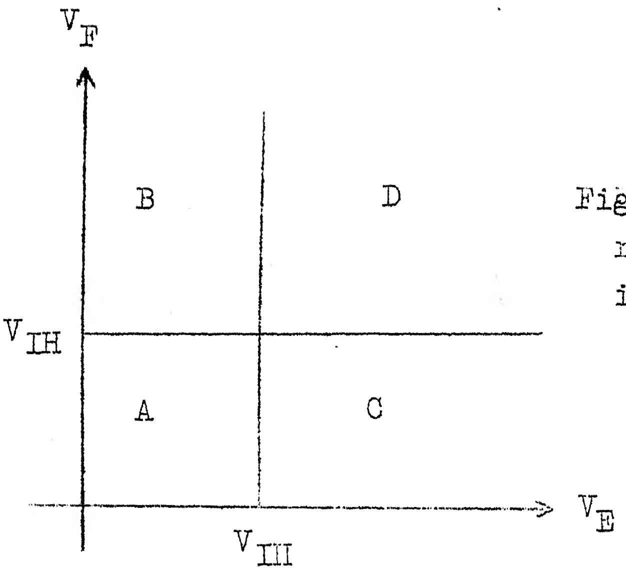


Figure 4.6.2(a): Orientation of regions when operating point is in region A.

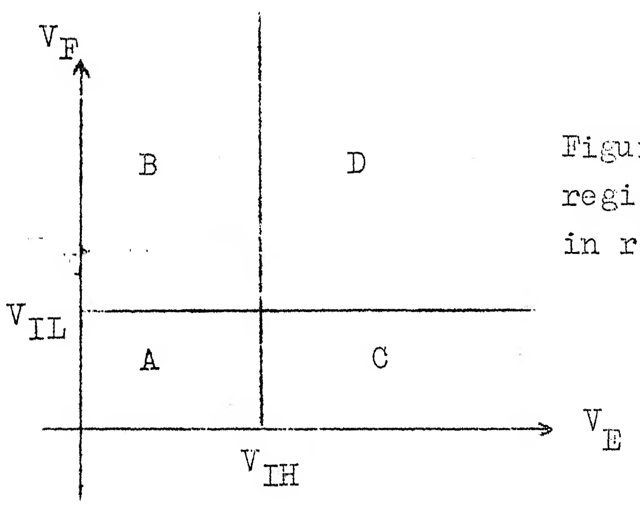


Figure 4.6.2(b): Orientation of regions when operating point is in region B.

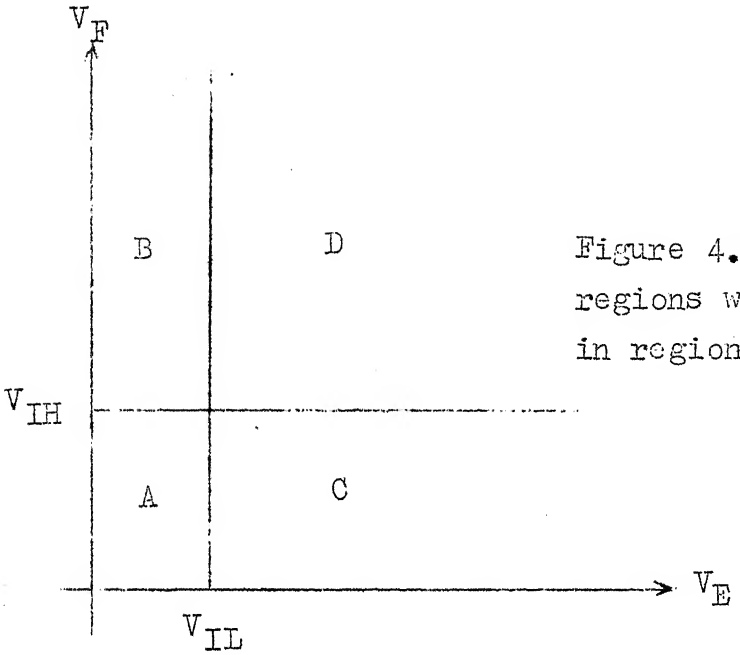


Figure 4.6.2(c): Orientation of regions when operating point is in region C.

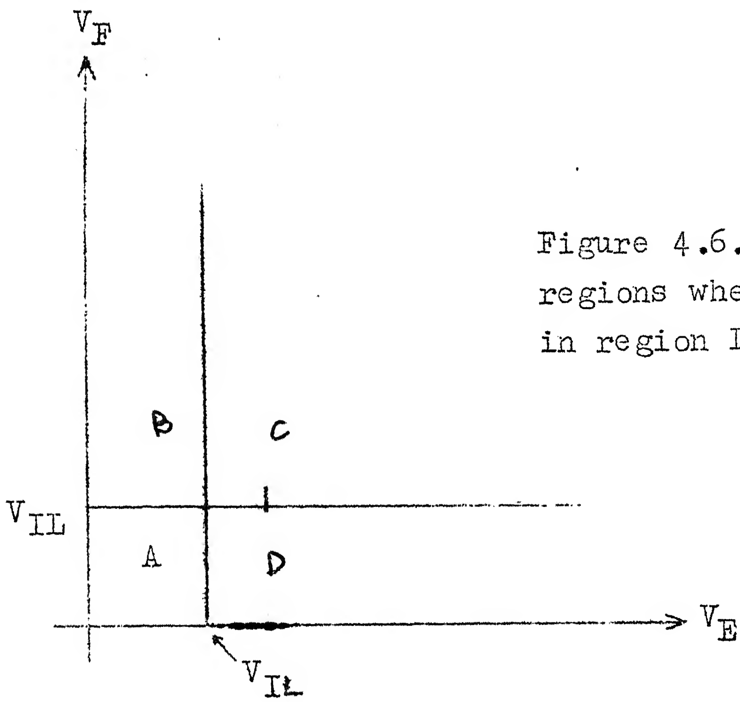


Figure 4.6.2(d): Orientation of regions when operating point is in region D.

This situation occurs when the hand is over strained in which case both the extensor and the flexor muscles are activated resulting in an ambiguity in the actuator drive. With these regions described, it is obvious that the actuator should stay idle when the operating point is in either of the regions A and D. In region B, the actuator is supposed to flex the fingers of the artificial hand, while in region C, it should extended them.

The transition regions are utilized to render the system insensitive to noise. The distribution of these transition regions to achieve noise immunity then, defines the effective operative region. The effective transition region over which the existing state of the actuator is maintained consists of the normal operative region along with the adjacent transition regions as indicated in Figure 4.6.2.

The scheme described above can be easily achieved by means of two hysteresis comparators using opamps. The circuit and the input output characteristic of such a comparator is shown in Figure 4.6.3. The scheme can be realized using two such comparators from a consideration of the four regions of operation as given in Table 4.6.1.

Table 4.6.1

	Region	V_{OE}	V_{OF}	$V_{OE}-V_{OF}$	Function of the actuator
(i)	A	V_{OH}	V_{OH}	0	Idle
(ii)	B	V_{OH}	V_{OL}	$(V_{OH} - V_{OL})$	Flex
(iii)	C	V_{OL}	V_{OH}	$-(V_{OH}-V_{OL})$	Extend
(iv)	D	V_{OL}	V_{OL}	0	Idle

Thus, our purpose is achieved if the voltage $V_a = (V_{OE}-V_{OF})$ is used to drive the actuator such that the direction of motion depends on the signal V_a. If, from the output of the comparators, V_a is to be obtained, a difference amplifier would be needed. This gives rise to problems due to offsets and CMRR. This can be avoided if one of the two signals V_{IE} and V_{IF} is inverted and fed to a negative voltage comparator, the other being fed to a positive voltage comparator and the comparator outputs added by a simple resistive summing as in Figure 4.6.4.are given by

The thresholds for the comparators shown in Figure 4.6.4 are given by

$$V_{IL} = (-\Delta V + V_T) \quad \text{and} \quad V_{IH} = (\Delta V + V_T)$$

for positive voltage comparator and

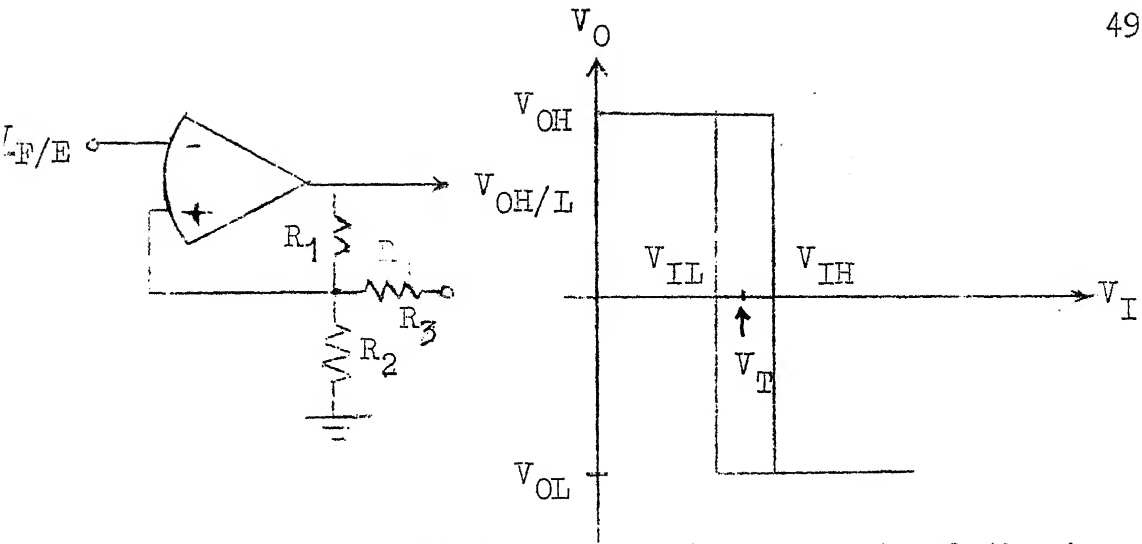
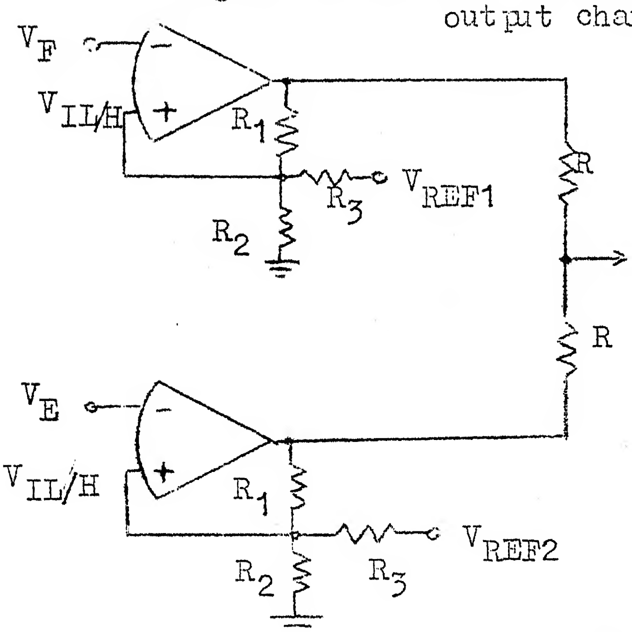
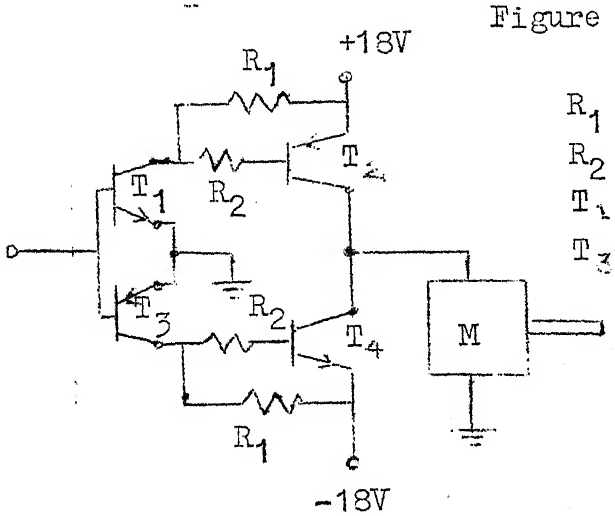


Figure 4.6.3: The comparator circuit and the input-output characteristic.



V_{REF1} = +9 Volts
V_{REF2} = -9 Volts
R_1 = 15 K
R_2 = 2.2K
R_3 = 3.3K
V_T = 3.3 Volts
ΔV = 0.72 Volts
R = 10K

Figure 4.6.4: The comparator stage.



R_1 = 1K
R_2 = 2.2K
T_1 = BC 148B
T_3 = BC158B
T_2 = SK100
T_4 = SL100

Figure 4.7.1: The driver stage.

$$V_{IL} = (+\Delta V - V_T) \quad \text{and} \quad V_{IH} = (-\Delta V - V_T)$$

for the negative voltage comparator where

$$V_T = \frac{R_2R_1}{R_1R_2+R_1R_3+R_2R_3}\left|V_{REF}\right| \; ; \; \Delta V = \frac{R_2R_1}{R_1R_2+R_2R_3+R_1R_3}\left|V_o\right|$$

The values of R_1, R_2 and R_3 arrived at, to obtain

$$V_{IL} = 2.6V \quad \text{and} \quad V_{IH} = 4.0V$$

are shown in the figure.

4.7 Design of the driver stage

The circuit of the driver stage is shown in Figure 4.7.1. The transistors T_1 and T_3 are matched PNP and NPN pair and so are the power transistors T_2 and T_4. This stage is supposed to drive the d-c permanent magnet motor which in turn operates the actuator. The motor should have a high torque (to provide a grip with enough firmness).

The Transistors T_1 and T_3 have a beta of 200 and the Transistors T_2 and T_4 have a minimum beta of about 30. For an input of above 0.6V, T_1 and T_2 conduct and for an input below -0.6V, T_3 and T_4 conduct. For inputs between +0.6V and -0.6V, all the transistors remain off and the motor gets no input current, hence stays still.

The d-c motor used has a speed of 11,000 r.p.m. and a current ratings of about 200 mA on no load and about 300 mA under full load. The values of R_1 and R_2 selected in order to drive the transistors into saturation are indicated in the figure. To avoid coupling of spikes (produced by the motor) with the other units, a separate supply of +18V and -18V has been used.

CHAPTER 5

THE ACTUATOR

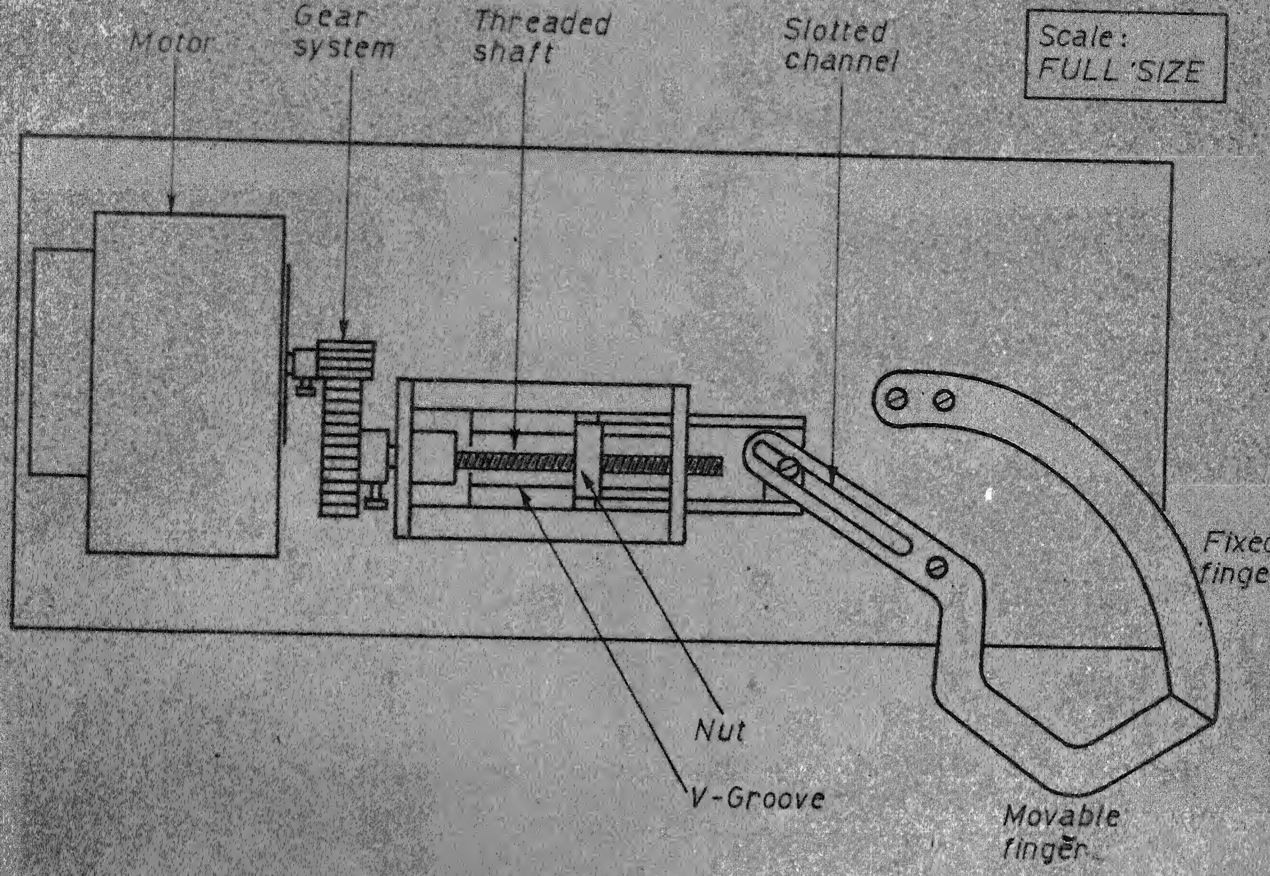
The actuator is the terminal device of any prosthetic system and includes a wide variety of artificial limbs to suit the needs of the amputees. The degrees of freedom presented by such devices depend upon the number of independent signals processed by the processor unit.

5.1 Design of the Actuator Unit

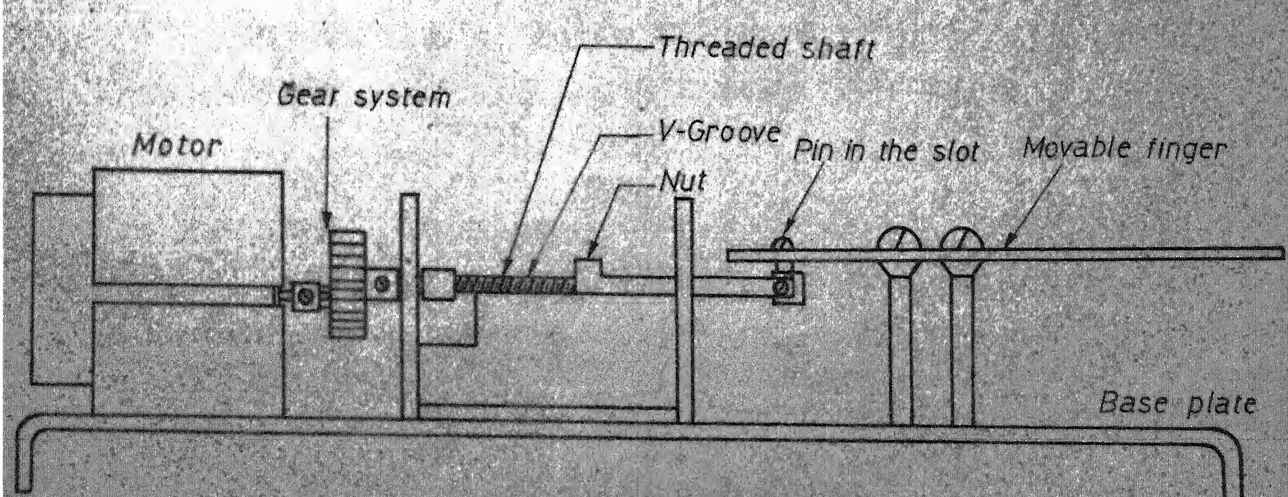
In this project, the aim has been to design the device to suit the needs of a below-elbow amputee. The signals of the human arm which could enable to flex or extend the artificial fingers of the prosthetic hand fitted on to the user, is used for this purpose.

A demonstration model of the mechanical actuator device that was designed to serve this purpose is shown in Figure 5.1.1. The motor has a speed of about 11,000 r.p.m. and a torque of about 0.011 H.P. Obviously, the torque offered is quite low and the speed, too high to make the flexing and extending of fingers look natural. Further, since the feedback of the performance is basically visual, it is desired to have the motion somewhat slower than normal. Hence speed reduction, associated with an increase in torque has been employed.

FIG. 5.1.1 THE ACTUATOR



PLAN



ELEVATION

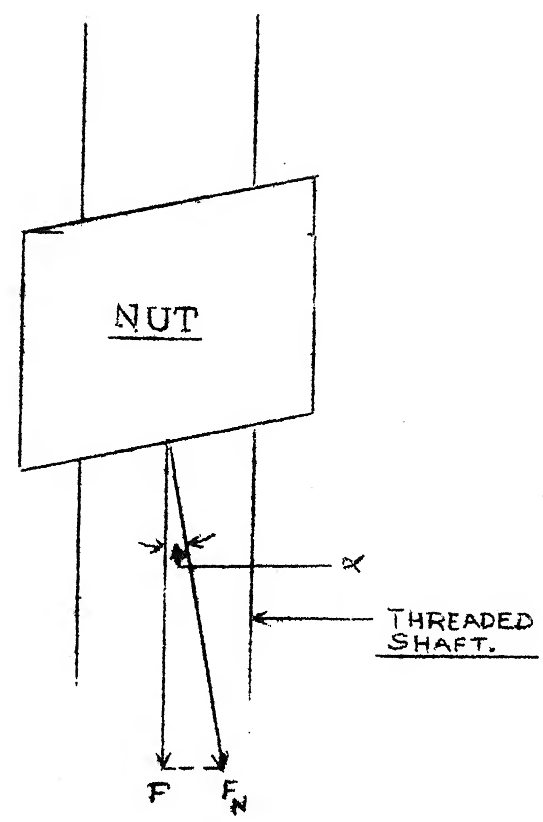


Figure 5.3.1: Force Diagram.

5.2 Operation of the Actuator

The operation of the actuator shown in the figure is as follows. By employing a gear system with a reduction ratio of 3, the threaded shaft is made to rotate at a speed one-third to speed of the motor. On the threaded shaft, a nut moves along the axis of the shaft depending on the direction of rotation of the shaft. This translatory movement of the nut is converted into angular movement of the artificial finger, by a pin which moves in a slotted channel made on the rear extension of the artificial finger mounted on a pivot. The screw is prevented from rotating around the axis of the shaft by providing a V-Groove in which it is allowed to slide along the axis of the shaft.

During the flexing movement, the rotation of the threaded shaft makes the nut slide along its axis in the groove towards the motor end. The movement of the pin closes the finger. When the finger comes to the extreme position, it is stopped from moving by a fixed finger, fixed on the base plate. If any object is placed in between, the two fingers grip the object with a certain force F. An equilibrium is reached due to the reaction of this force and the force exerted by the motor in an effort to pull the nut. If the motor is now switched off, the grip is maintained by frictional forces acting on the threads of the nut which prevents the nut from sliding along the threaded shaft.

When the motor shaft rotates in the other direction, the nut is made to slide forward and the fingers extend. In the extreme extended condition, the supports of the fixed finger prevent further sliding of the nut.

5.3 Estimation of the Torque of the Motor Required

The mechanism explained above is essentially that of a screw jack. If F is the gripping force, then, the torque T required of the motor is determined as follows:

Work done by the motor in rotating through an angle $\Delta\theta = \Gamma$

If p is thepitch of the screw, then, the distance moved by the nut for a rotation of $\Delta\theta$ by the motor shaft is

$$\Delta X = (\Delta\theta \, p/R)$$

where R is the reduction ratio of the gear system.

If Δy is the distance moved by the finger for rotation of motor shaft, then, work done on gripped object = $F\Delta y$.

Now if the lever is considered lossless,

$$F\Delta y = F_N \Delta X \, \mathrm{Cos}\, \alpha$$

where F_N is the force acting normal to the thread and α is the angle of the helix given by

$$\alpha = \tan^{-1}(p/2\pi r)$$

where r is the mean radius of the shaft.

Therefore,

$$F_N = F(\Delta y)/(\Delta X)\cos\alpha$$

If μ is the co-efficient of friction, then,

Loss of energy due to friction $= \mu\, F_N\, r(1/R)(\Delta\theta)$

Hence, by conservation of energy, we have:

$$T\Delta\theta = F\Delta y + \mu\, F_N\, r\, \frac{1}{R}\,\Delta\theta$$

$$\therefore \quad \Gamma = F\left(\frac{\Delta y}{\Delta\theta} + \frac{\mu(\Delta y)\, r}{(\Delta X)(\cos\alpha)R}\right)$$

The demonstration model designed has R = 3, p = 0.05 cm, r = 0.15 cm, $\Delta X/\Delta y = 1$, $\mu = 0.07$ (for mild steel contacts this value may be assumed for estimation purpose), and $\Delta y/\Delta\theta = 2.655\text{x}10^{-1}$. If a gripping force of 2 Kg is required at the tips of the fingers, a crude estimate of the torque of the motor under quasi-static condition is found to be about $9.12\text{x}10^{-4}$ Kg-m.

CHAPTER 6

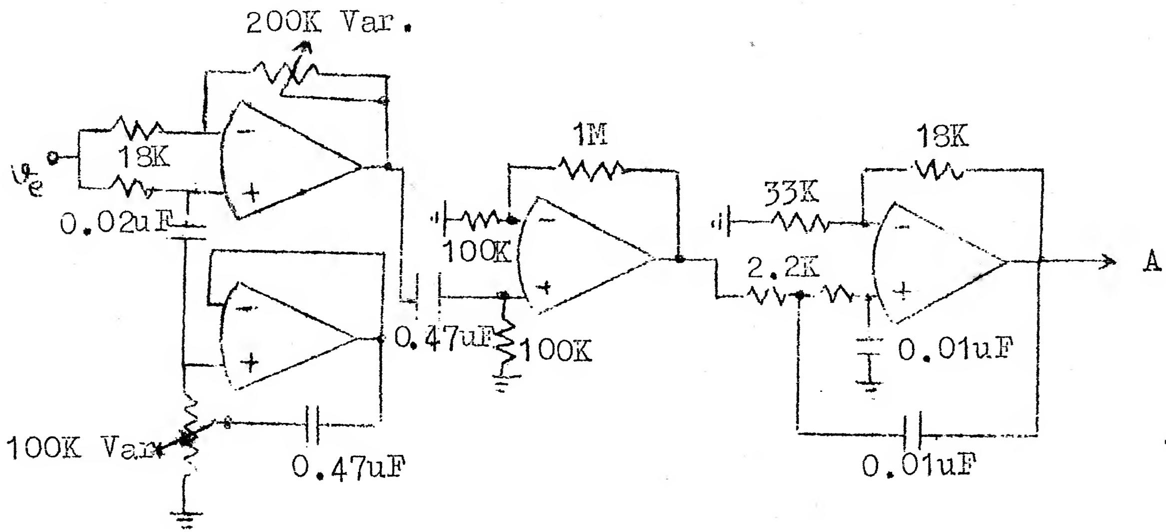
CONCLUSION

The various aspects of design have been covered in the preceeding chapters. In this chapter, we proceed to describe the details of fabrication of the control unit, directions for use and the test results. Certain modifications that might lead to an improvement over the fabricated system have been suggested.

6.1 Fabricated Unit

Two set of electrodes have been made on small elliptic printed cards and connected to the processor unit using flat flexible cables. Provision is made for rubber straps so as to strap the electrodes on the fore-arm.

The processor unit has been housed in a small aluminium case, the body being earthed. This unit has been fabricated on three printed cards. Card 1 consists of the preprocessor section of the processing unit of the two channels. Card 2 consists of the Final processing section of one of the channels and the negative voltage comparator. Card 3 has the final processing section of the other channel and the positive voltage comparator. The circuit on each of these three cards is shown in Figure 6.1.1(a) - (c). The



Points A and B are on CARD 2.

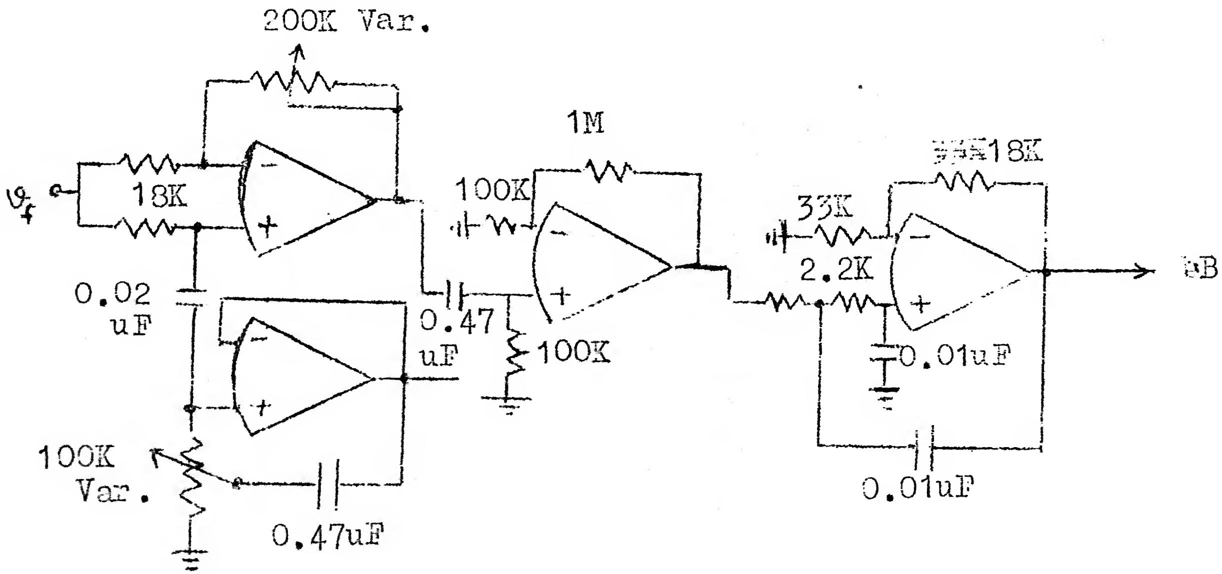


Figure 6.1.1(a): CARD 1.

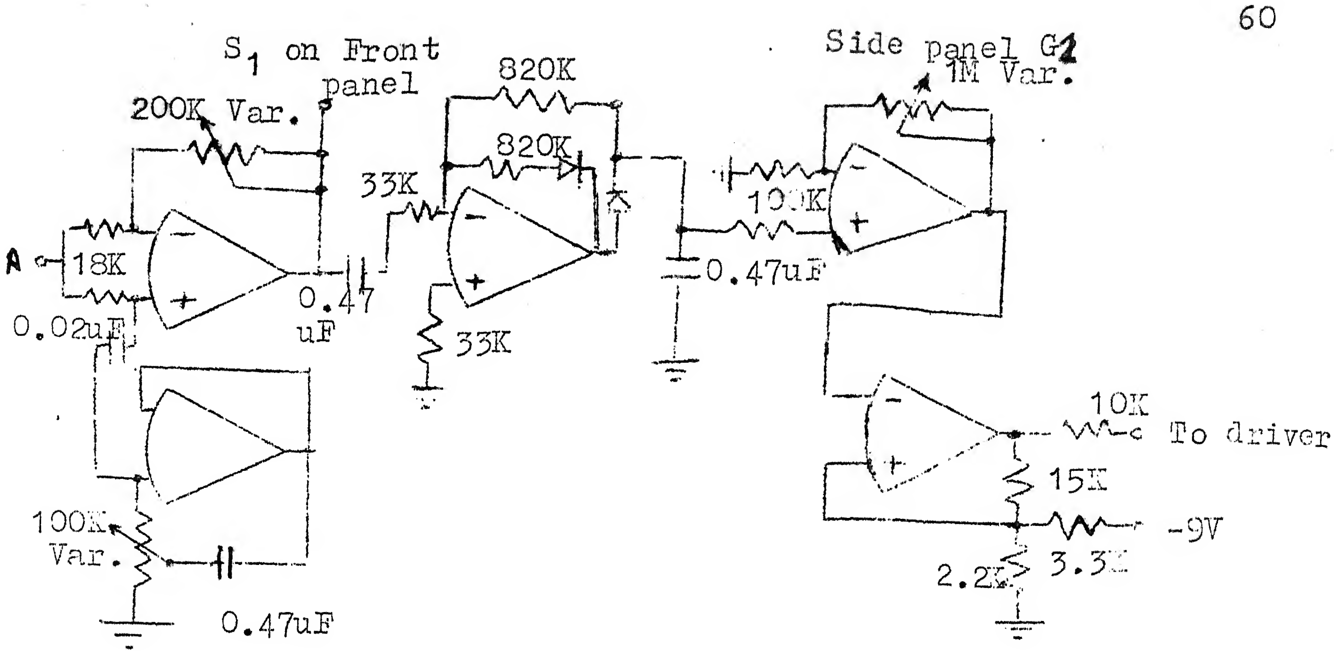


Figure 6.1.1(b): CARD 2

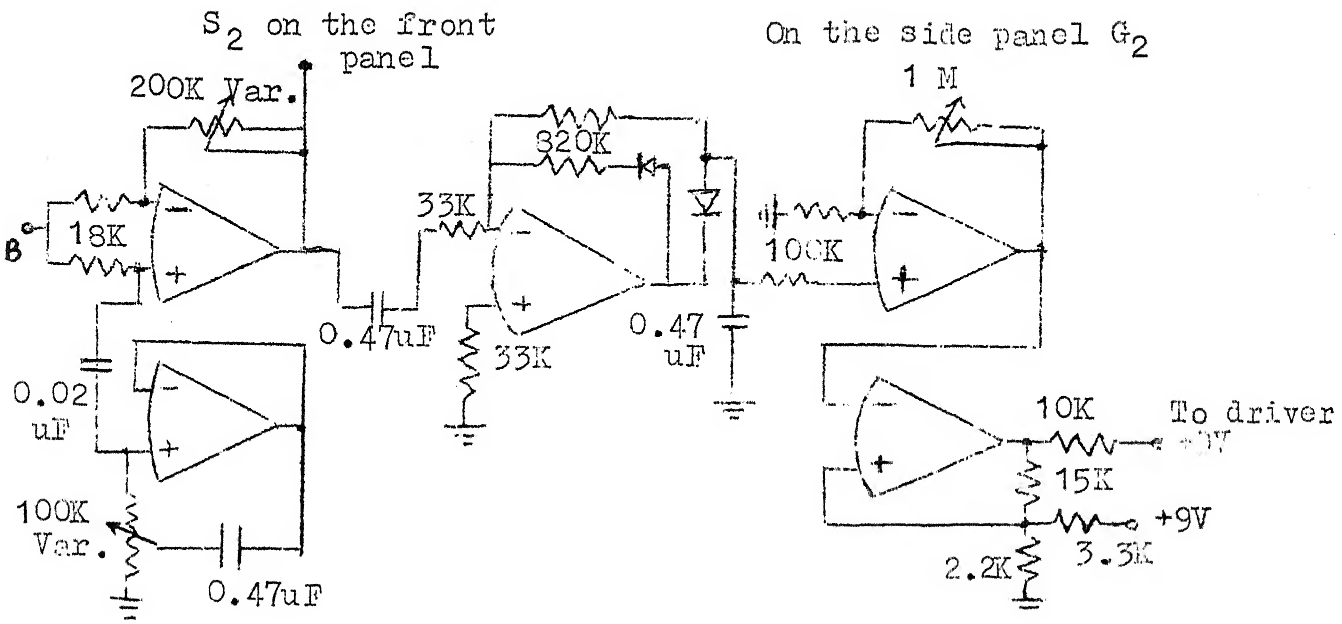


Figure 6.1.1(c) Card 3.

driver stage which is on card 4 is mounted on the actuator itself.

The gain controls for the two channels are provided on the side panel. The preprocessor outputs from the two channels are provided on the front panel to enable observation of the signals on a CRO. Signals observed at this point have an amplitude of a few millivolts. The final comparator output is brought out on the front panel, and is externally connected to the driver stage by means of flexible wires. The system requires $\pm 9V$ power supplies, which have to be externally connected to the terminals provided on the front panel.

6.2 Directions for Use

Before the electrode is fitted on, care has to be taken to see that the skin surface is clean. Before the electrode is strapped on, they should be placed on the site and waveforms observed on the CRO in order to select the most appropriate site. Then, the electrodes are strapped on and adjusted such that the three studs make proper contact with the skin. Once this is done, the potentio meters provided on the channels must be adjusted such that when voluntary contraction of muscle is carried out, the corresponding signals amplify to the required extent and provide the proper inputs to the comparator.

After these adjustments, the processor unit is connected to the driver stage and the driver stage is switched ON.

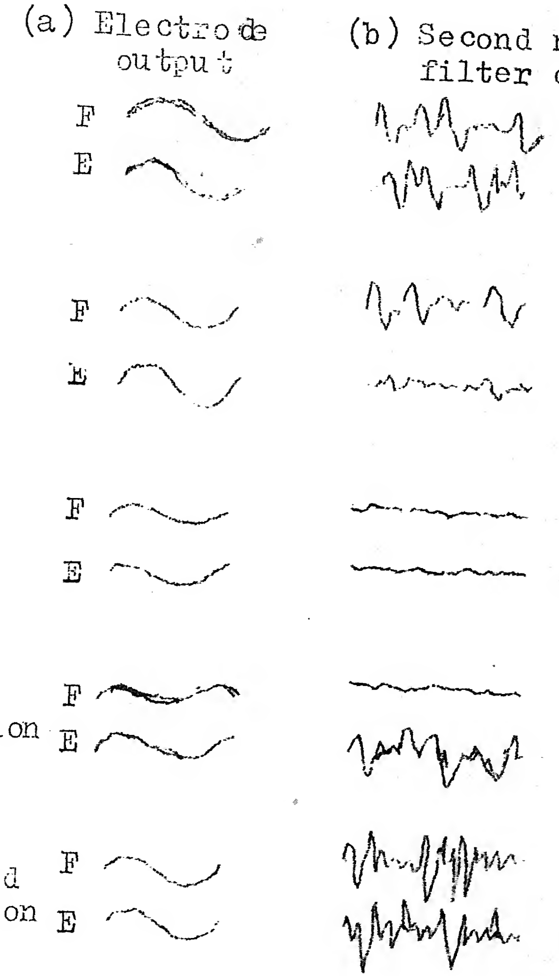
6.3 Test Results Obtained

In the experiments carried out on a normal hand, the waveforms observed at various points in the circuit for different stances of the hand are indicated, in Figure 6.3.1. It was noted that the art of activating the desired group of muscles such that the desired function is performed by the actuator could be learnt with a fair amount of ease.

The signal output of each electrode was of the order of 100 to 300 μV. Movement artefacts were found to be the main cause of disturbance. Partial activities of the extensor and the flexor muscles during rotation of the wrist also were found to be the cause of malfunctioning. This may be eliminated by designing the electrode to have the three contact studs closer such that the studs can be concentrated close to the required group of muscle fibres, thus eliminating noise pickup due to the activity of other muscles.

Hands with thicker layer of fat were seen to produce lesser signals. This is due to the fact that the signals get attenuated as they pass through the layers of fat.

The system was tried on various individuals and found to work satisfactorily.



Condition	Channel	Final processor output	Comparator output	Driver output
Strained flexion	F	-8 to -10V d.c.	+12V	0V
	E	+7 to +9V d.c.	-12V	
Normal flexion	F	-7 to -9V d.c.	+12V	+12V
	E	+3 to +4V d.c.	+12V	
Relaxed hand	F	-3 to -4V d.c.	-12V	0V
	E	+3 to +4V d.c.	+12V	
Normal extension	F	-3 to -4V d.c.	-12V	-12V
	E	+7 to +9V d.c.	-12V	
Strained extension	F	-7 to -9V d.c.	+12V	0V
	E	+7 to +9V d.c.	-12V	

F: Flexor channel; E: Extensor channel. Scale: (a) X:1 cm=5V
Y:1 cm=0.01 sec.
(b) X:1 cm=100 mV
Y:1 cm=0.01 sec.

Figure 6.3.1: Waveforms observed at various points in the system.

with rubber gloves so as to give a natural appearance. The actuator system may be designed on the principle of slipping clutch mechanism.

Feedback may be provided from the finger tips such that when any object has been gripped with a predetermined force, the motor is stopped.

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